

Quantification of Cumulative Load on the Knee Using a Vibration
Emission Method

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ABSTRACT

Background: Squat-lifting postures can induce heavy mechanical loads on the joint, causing the articulating surfaces to deform. This can result in changes in the vibration characteristics of the joint surfaces. The purpose of this study was to examine the influence of cumulative mechanical load on the vibration properties of the knee joint and to gain an understanding of how these properties may relate to an increase in cumulative load placed on the joint.

Methods: In this study, cumulative load was measured as the summation of knee joint torque during squat lifting, while a piezoelectric accelerometer was used to capture vibration signals from points on the knee during flexion and extension. Twelve university students were recruited for a repeated measures study. Each participant attended one session where they had to perform a series of six squat-lifting tasks on a force platform. Motion capture equipment was used to obtain kinematic data. The cumulative 3-D moment on the joint was calculated using inverse dynamics.

Results: Statistical analyses by a t-test and ANOVA showed a decreasing trend in the RMS power of the signal captured in the 750-Hz-2000Hz range before and after mechanical load was induced by squat lifting. A linear regression analysis indicated a significant correlation between cumulative 3-D moment on the knee joint and the median frequency of vibration signals from the patella during flexion in the 1000 Hz - 2500 Hz range.

Conclusions: The results of this study indicate the possibility of a relationship between mechanical exposure on the knee joint and its vibration properties during joint movement. Despite the small sample size, a declining trend was observed in the normalized RMS power of signals with increase in loading. However, the quantitative nature of this relationship is not clear and the current study points towards a possible multi-variate or a non-linear relationship between joint exposure and knee vibrations. Future studies must investigate these possibilities using direct measures of joint loading, cartilage deformation and their relation to joint vibrations.

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TABLE OF CONTENTS

1. INTRODUCTION	1
1.1 Rationale.....	1
1.1.1 The problem of occupational knee joint trauma.....	1
1.1.2 Restricted workspaces and joint deformation.....	2
1.2 Specific aims and hypotheses.....	5
1.3 Implications.....	6
2.0 LITERATURE REVIEW	8
2.1 Anatomy of the knee	8
2.1.1 Parts of the knee joint	8
2.1.2 Locking mechanism.....	9
2.1.3 Diseases of the knee	9
2.2 Etiology and mechanics of cartilage degeneration.....	10
2.3 Cumulative exposure and cartilage degeneration.....	13
2.3.1 Cumulative exposure	13
2.3.2 Measuring cumulative exposure on joints	15
2.4 Introduction to non-destructive testing methods and vibration emission	17
2.5 Vibration emission in joints	18
2.5.1 Physics behind vibration emissions.....	18
2.5.2 History of joint vibration emission.....	18
3.0 METHODS	21
3.1 Participants	21
3.2 Instrumentation.....	21
3.2.1 Motion capture system.....	21

3.2.2 Accelerometer.....	22
3.2.3 Metronome.....	23
3.3 Pilot study.....	23
3.3.1 Procedure	23
3.3.2 Dependent variables and predictor variables	24
3.3.3 Data analyses	25
3.3.4 Pilot results	25
3.3.5 Issues with pilot and revisions made	28
3.4 Final experimental procedure.....	28
3.5 Data analysis	32
3.5.1 Joint moment calculation (mechanical loading).....	32
3.5.2 Joint vibration signal processing	36
3.6 Statistical analysis	40
4.0 RESULTS	42
4.1 Analysis from ensemble averages	42
4.2 Regression analyses.....	46
4.3 Cumulative moment normalized with respect to body mass.....	52
4.4 Two-factor regression model	52
5.0 DISCUSSION	55
5.1 General findings	55
5.2 Frequency range under consideration	56
5.3 Individual differences in vibration behavior due to diurnal changes and time-dependent joint deformation.....	57
5.4 Only the signals captured from the patella during passive movement of limb were consequential.....	58
5.5 Possibility of a non-linear or multi-variate relationship.....	59

6.0 LIMITATIONS AND FUTURE DIRECTIONS	60
6.1 Etiology of vibrations.....	60
6.2 Direct measures of loading.....	61
6.3 Static versus dynamic loading.....	62
6.4 Advanced signal processing methods	63
7.0 CONCLUSION	64
REFERENCES.....	65
APPENDIX A: DEMOGRAPHIC QUESTIONNAIRE	71
APPENDIX B : CONSENT FORM	72

LIST OF FIGURES

Figure 3.1: Accelerometer unit used in the study -----	22
Figure 3.2 (a): Relative root mean square (RMS) power of the signals from the patella measured after each task. The RMS power here is normalized to its initial value computed from the signals which were recorded at the start of the experiment, before the participant performed any tasks. -----	26
Figure 3.2 (b): Relative root mean square RMS power of the signals from the medial condyle measured after each task. The RMS power here is normalized to its initial value computed from the signals which were recorded at the start of the experiment, before the participant performed any tasks. -----	27
Figure 3.2(c): Relative Median Frequency (RMF) of signals measured from the medial condyle after each task. The median frequency is normalized with respect to its initial value which is computed from the signals recorded at the start of the experiment, before the participant performs any tasks. -----	27
Figure 3.2(d): Relative Median Frequency (RMF) of signals measured from the patella after each task. The median frequency is normalized with respect to its initial value which is computed from the signals recorded at the start of the experiment, before the participant performs any tasks. -----	28
Figure 3.3: (a) Accelerometer positioned on the mid-patella. (b) Accelerometer positioned on the medial condyle. -----	29
Figure 3.4: Experimental set-up for performing squat-lifting tasks in (a) Resting position and (b) Half-Squat position. -----	31
Figure 3.5: Free body diagram showing forces and moments acting on the (a) Foot segment and (b) Shank segment -----	34
Figure 3.6: Flowchart depicting steps involved in processing vibration data. -----	39
Figure 4.1(a): Ensemble average of the Fourier transforms for signals from the medial condyle during extension. -----	42
Figure 4.1(b): Ensemble average of the Fourier transforms for signals from the medial condyle during flexion -----	43
Figure 4.1(c): Ensemble average of the Fourier transforms for signals from the patella during extension -----	43
Figure 4.1(d): Ensemble average of the Fourier transforms for signals from the patella during flexion -----	44

Figure 4.2: Graph displaying the mean and standard deviation of the relative RMS power across all participants, at different points in time based on the number of tasks accomplished. The decrease in the mean relative RMS power is apparent after the participants accomplish two tasks and all four tasks. ----- 46

Figure 4.3 (a) : Regression plot of Cumulative moments versus Relative Median Frequency for vibration signals from the Patella in the range 1000-1500Hz during flexion showing individual scatter of points for each participant. Each unique combination of color and shape represents a participant. ----- 48

Figure 4.3 (b): Residual plot of linear regression between 3-D cumulative moment and RMF, showing a random, but non-normal scatter of points. ----- 49

Figure 4.3 (c): Regression plot of the Log of 3-D Cumulative Moment at the knee versus the Relative Median Frequency RMF for 1000-2500Hz range. The coefficient of determination R^2 is now 33% as opposed to 30% when the regression was performed without the logarithm being taken.----- 49

Figure 4.4 (a): Regression plot of Cumulative moments versus Relative Median Frequency for vibration signals from the Patella in the range 20Hz-100Hz during flexion showing individual scatter of points for each participant. Each unique combination of color and shape represents a participant. ----- 50

Figure 4.4 (b): The residual scatter plot for the linear regression performed between 3-D cumulative load and the RMF for vibration signals in the 20Hz-100Hz range recorded from the patella during knee flexion.----- 51

Figure 4.4 (c): The regression plot of the log of 3-D cumulative moment versus vibration signal RMF in the range 20Hz-100Hz captured from the patella during knee flexion.----- 51

Figure 4.5: Scatterplot of actual versus predicted values of the cumulative moment normalized to body mass. ----- 53

LIST OF TABLES

Table 3.1: Tasks assigned based on combinations of load and frequency of lifts. -----	24
Table 3.2 (a): Sample RMS values obtained for the patella. -----	26
Table 3.2 (b): Sample Median Frequency values obtained for the patella. -----	26
Table 3.3: Counterbalancing scheme and description of tasks assigned to each individual. ----	32
Table 4.1(a): RMS values for vibration signals in the range 750Hz-2000Hz, from the patella during flexion. -----	45
Table 4.1(b): Median Frequency values for vibration signals in the range 750Hz-2000Hz, from the patella during flexion. -----	45
Table 4.2: Results of all regressions performed for Cumulative Load (Nm.s) versus RMF. * indicates statistical significance. -----	47
Table 4.3 (a): Mean median frequency (with Std. Dev) computed for signals recorded after each task from the patella, while under knee flexion. -----	47
Table 4.3 (b): Mean RMS power (with Std. Dev) computed for signals recorded after each task from the medial condyle, while under knee flexion. -----	48

1. INTRODUCTION

1.1 Rationale

1.1.1 The problem of occupational knee joint trauma

The diagnosis of musculoskeletal injuries dates back to the 17th century, when Ramazzini attributed chronic joint pain in clerks and scribes to repetitive motions (Kroemer, 1989). During the 18th and 19th centuries, the industrial revolution witnessed the advent of several transformations in work design and material handling (Dempsey and Mathiassen, 2006). Labor was divided and hours of work increased as manufacturing processes had to catch up increasing demands of goods. Workplace design philosophies were based on overall efficiency of producing an output rather than the workers' individual efficiency of labor. These transformations introduced restricted workspaces which placed various physical demands on workers, such as joint stresses on the knee due to constrained postures. Even today, BLS (2010) statistics across private and public ownerships suggest that musculoskeletal injuries of the knee joints contribute to a median of 15 days away from work, second only to shoulder injuries. In an analysis of mining injury claims spanning an average of 7.8 years, Gallagher and Moore *et al.* (2009) ranked knee injuries in the mining industry to be the costliest and most frequent of all body parts, with a total cost of \$4,172,600 (May 2008) and 318 occurrences. They also estimated the average claim for each knee injury to be \$13,100 resulting in knee claims contributing to 17.39% of the total injury cost in mining. These numbers highlight the need to address the problem of musculoskeletal trauma in the knees due to occupational factors such as repetitive motion and constrained spaces.

Furthermore, productivity could become sub-optimal if work related joint injuries were not taken into consideration while keeping up with progressive trends such as Just-In-Time manufacturing (Kuorinka 1998). Martimo *et al.* (2009) investigated productivity self-reports of employees with upper extremity musculoskeletal disorders in the computer industry and discovered that the productivity loss associated with pain and its interference in work was substantial. A recent study has attested work productivity itself as an important variable measure in assessing the prevalence of musculoskeletal disorders in the workplace (Escorpizo, 2008).

Musculoskeletal trauma can be acute or cumulative. Acute trauma is associated with acute injuries manifesting as fractures or sharp sudden pain. Cumulative trauma is the result of repetitive, sustained or forceful motions exerted over a long period of time and can often manifest as chronic joint pain (Kroemer 1989). Exploring and quantifying risk factors associated with cumulative trauma on different parts of the body has been an area of growing interest. Kumar (1990) and Norman et al. (1996) were among the first to quantify cumulative exposure on the low back among industry workers. More recently, Bosch et al. (2007) attempted to quantify perceived discomfort and muscle fatigue developed over time while performing low intensity repetitive assembly tasks and found a temporal increase in both characteristics. Gregory and Callaghan (2008) examined participants' physiological responses while they performed light precision tasks in a prolonged standing posture and concluded that there is a propensity to develop discomfort in the low back while standing and working for extended time periods.

Ergonomic technologies developed for counteracting low back pain among miners showed a 10% drop in cumulative low back pain from 1984 to 2004. However, a corresponding 10% rise in the incidence of cumulative knee injury was also reported at the same time (Moore, 2008). This implies that injury risk did not decrease per se, but that the physical demands may have simply been placed on a new area of the body. Therefore, knee joint trauma cannot be ignored today in terms of compensation cost and the number of incidences. Lately, there has been a growing interest in quantifying cumulative load on the knee. Studies have used the knee adduction moment during walking as a proxy for estimating the total exposure on the knee joint (Robbins et al., 2009). However, more research is needed regarding the effects of cumulative exposure on the knees due to occupational factors such as static joint stresses due to constrained postures, dynamic loading, and frequency of lifting.

1.1.2 Restricted workspaces and joint deformation.

Several occupations within the agriculture, forestry, mining, production, and construction industries require workers to adopt uncommon postures such as squatting and kneeling. Workers' position and motion has been associated with 13.9% of the total cases of non-fatal injuries and illnesses in 2009 (BLS, 2010). Much of the ergonomic risk in these occupations is attributed to the nature of the job and the restricted work space. Such jobs can result in high mechanical loads on the knee as the worker may have to kneel or squat all day long. Often times

heavy lifting and carrying is involved along with squatting. Researchers have reported that occupations involving such postures are associated with an increased risk of knee osteoarthritis. A dose-response relationship was reported between the time of exposure in these uncommon postures and the risk of disease (Seidler, Bolm-Audorff et al. 2008; Klusmann, Gebhardt et al. 2010). The association was found to be strongly elevated during combined exposures where lifting or carrying of weights was performed in kneeling or squatting postures (Cooper, McAlindon et al. 1994; Coggon, Croft et al. 2000; Seidler, Bolm-Audorff et al. 2008). Furthermore, knee joint moments reach peak values in deep knee flexion (Nagura, Dyrby et al. 2002). Squatting and knee bends in normal joints revealed a significant increase in cartilage deformation over neutral postures as viewed under MRI scans (Eckstein, Lemberger et al. 2005).

The results of these epidemiological observations question the adequacy of the articular cartilage of the knee to support the loads it is subjected to over time in these occupations. The incidence rate of knee disorders cannot be expected to decline without efficient exposure assessment tools. Previous studies which have attempted to quantify occupational exposure on the knee joint have been based on telephonic interviews and self-report questionnaires (Seidler, Bolm-Audorff et al. 2008; Mannien, Heliövaara et al., 2002). These studies have reported a higher risk of chronic arthritis among occupations which require regular squatting and stair climbing over those which don't. The disadvantage with these self-report methods is that the exposure assessment reflects the outcome of years of exposure. A more proactive approach would be to directly examine cartilage health for lesions or deformation. MRI scans have been used in the past to assess the *in vivo* deformation of the cartilage but these methods require elaborate equipment and cartilage health assessments are impractical in field environments (Eckstein, Lemberger et al., 2005). Hence, industries such as mining need to focus on improving methods of assessing cumulative knee trauma due to uncommon postures. An alternative, less explored non-destructive method for assessing cartilage health could involve use of piezoelectric transducers to capture vibrations from the viscoelastic cartilage. Use of these transducers for exposure assessments is worth exploring owing to the portability and ease of use of these instruments.

Vibration emission methods have been used in the past, to determine the functional integrity of implants (Georgiou and Cunningham 2001) and to distinguish between normal and pathological knees *in vivo* (Tavathia, Rangayyan et al. 1992; Mascaro, Prior et al. 2009). In these methods, a piezoelectric accelerometer attached to bony prominences on the knees was used to capture vibrations from a “loose” or diseased knee. These vibration signals were sensitive to local vibrations and can detect micro-damage in the cartilage. A deformed cartilage would emit viscoelastic vibrations *in vivo* thereby releasing energy. The advantage these methods hold is that the equipment is portable and so, data could be collected on the field without the need for laboratory tests. Besides this, there is scope for proactively determining the health of the cartilage. This could enable keeping a check on workers’ joint health while administering controls for safe working conditions accordingly.

As discussed earlier, compensation costs due to workplace musculoskeletal injury can be extremely expensive for companies. In order to avoid such costs and to ensure safe working conditions, the industry is interested in determining exposure limits and measuring joint exposure so that pain reports can be proactively predicted. Research directed towards quantifying lumbar mechanical exposure has emphasized on the value of cumulative joint load as an important risk factor in epidemiologically distinguishing cases from controls (Kumar, 2000; Norman et al., 1998). Maly (2008) asserted the need for an integrated loading estimate for the knees as the long term effects of repetitive stress on joint tissues cannot be ignored when accounting for joint pathologies. She also suggested the use of the knee adduction moment as a suitable proxy for estimating joint load. Cumulative daily exposure on the knees has been quantified in terms of the knee adduction moment and tested for its reliability (Robbins et al., 2009). However, this moment was specific to walking only. In the current study, participants performed several squat-lifting tasks. The kinetics of squat lifting can be complex when compared to walking as it may not be restricted to the sagittal plane. Therefore in this study, the 3-D joint moment at the knee, integrated over the task duration is used as a measure of exposure.

1.2 Specific aims and hypotheses

The overarching goal of this study was to test the use of a vibration emission technique for investigating the effect of squat-lifting on vibrations from the articulating surfaces of the knee. Since viscoelastic deformation of the articular cartilage is influenced by its mechanical loading environment, the vibration signal of the knee under load might differ from its resting condition. Therefore, our first specific aim and corresponding hypotheses were as follows:

SPECIFIC AIM 1: To test if there are significant differences in vibration characteristics of the articular cartilage before and after loading the knee joint.

The characteristics considered for investigation in the present study included Root Mean Square power (RMS) of the vibration signal in time domain and the median frequency of the signal in frequency domain. The average power of this vibration signal was found to increase with rising levels of pathology in the frequency bandwidth of 100-500Hz (Reddy, Rothschild et al., 1995). As cartilage deformation in normal joints increases with mechanical loading, it is expected that energy dissipation will also increase. So, the first hypothesis is:

Hypothesis 1a:

H₀: There is no difference in the RMS of the signal before and after loading the knee.

H_{a1}: Power of the acoustic signal from the knee joint, expressed as the RMS of the signal, is expected to increase with increase in mechanical loading at the knee.

The frequency range in which these signals were emitted was found to decline as the pathologies worsened (McCoy, Mollan et al., 1987). Therefore, normal joints exhibited vibrations in higher frequencies compared to pathological joints. Thus, the severity of cartilage deformation is greater as the signal frequency declines and this is the basis of our next hypothesis.

Hypothesis 1b:

H₀: There is no effect of loading on the median frequency of the signal.

H_{b1}: The median frequency of the vibration signal is expected to decrease with increase in mechanical loading at the knee.

If these differences exist across loading conditions, it may be possible to quantify cumulative load on the joint in terms of the vibration properties measured. To this end, our second aim and hypothesis were as follows.

SPECIFIC AIM 2: To build a model of cumulative load versus vibration emission characteristics for the sake of quantifying the former in terms of the latter.

The characteristic could be RMS power of the signal or the median frequency, based whether or not significant differences exist in vibration signal characteristics across loading conditions. As a first step towards discovering a relationship between mechanical load and the vibration signal, we hypothesize that:

Hypothesis 2a:

H₀: There is no relationship between the 3-D cumulative moment at the knee and the RMS of the knee vibration signal after loading the joint.

H_{b2}: The 3-D cumulative moment on the knee joint varies linearly with the median frequency of the vibration signal recorded from the joint.

Hypothesis 2b:

H₀: There is no relationship between the 3-D cumulative moment on the knee joint and the median frequency of the vibration signal.

H_{a2}: The 3-D cumulative moment on the knee varies linearly with the RMS of the vibration signal recorded from the knee.

1.3 Implications

- The piezoelectric transducer is an easy to use and portable device. Data collection using these devices can be performed *in-vivo* and on-site. This could potentially enable new, field environments for conducting experiments with direct measurement equipment. Experiments conducted in laboratories come with limitations such as a requirement of elaborate equipment, smaller sampler sizes, and generalizability of results to wider populations.

- Use of vibration emission proactively to assess cartilage health could act as a preventive measure for administering controls to improve workplace mechanical environment.
- Expression of vibrations in terms of mechanical load and frequency can help establish guidelines for safe limits of loading and frequency of lifting, based on cartilage health assessments.

2.0 LITERATURE REVIEW

2.1 Anatomy of the knee

The knee joint is the largest joint that performs functions of movement, such as weight bearing and smooth articulation of joints. Such joints are called synovial joints. The knee is built around two articulations. The first is the articulation between the femur and the tibia; the other is between the patella and the femur. The following topics extracted from Gray's Anatomy for Students (Richard L. Drake, Wayne Vogl et al., 2005), are a brief review on the anatomy and important functions of the knee.

2.1.1 Parts of the knee joint

Ligaments: The major ligaments in the knee are the patellar ligament, the medial and lateral collateral ligaments and the anterior and posterior cruciate ligaments. The collateral ligaments are located approximately parallel to each other on either side of the knee joint. They provide reinforcement and play a major role in stabilizing the hinge-like movement of the joint. The cruciate ligaments cross each other in the sagittal plane between the femoral and tibial surfaces through the intercondylar space. These ligaments are essential for supporting joint movements and preventing displacement of the articulating parts over each other. Rupture or injury in the cruciate ligaments can result in abnormal loads on the articulations. The cruciate ligaments are located outside the articular cavity, but are enclosed within the fibrous membrane of the knee joint. The patellar ligament connects the apex of the patella to the tibial tuberosity.

Articular surfaces: These form the surfaces of the bones at which the joint meets. They are covered by the hyaline cartilage and participate in movement. The most important surfaces include the two femoral condyles and the adjacent surfaces of the superior aspect of the tibial condyles. The femoral condyles are round in shape during full flexion of the knee and are flat during full extension. All joint surfaces are enclosed within a single articular cavity.

Menisci: The knee joint contains two C-shaped fibro cartilaginous menisci, on the medial and lateral side. The menisci function as a cushion to the articulating condyles, such that the congruency of the joint is protected during movement.

Synovial membrane: This membrane is found at the margins of the articular surfaces and extends into a recess and a bursa. Its function is to provide low friction surfaces for tendon movement associated with the joint. It is separated from the patellar tendon by a layer of infrapatellar fat.

Fibrous membrane: This is an extensive membrane enclosing the articular cavity and the intercondylar region. It is largely made up of reinforcements and extensions of surrounding muscles and tendons.

Vascular supply: Blood vessels from the superior and inferior parts relative to the knee joint combine to form an anastomotic network around the joint. The joint is innervated by branches from the obturator, femoral, tibial and fibular nerves.

2.1.2 Locking mechanism

The amount of work required by the joint to maintain posture is optimized. The locking mechanism contributes to this function in two ways. Firstly, the shape and size of the femoral condyles changes with flexion and extension. In flexion, the surfaces are rounded in the posterior aspects of the femoral condyles. As the knee extends, the surfaces move to the broad and flat region on the inferior part of the condyles. Thus, the joint surfaces become larger and more stable in extension. Secondly, medial rotation of the femur on the tibia during complete extension tightens all the associated ligaments and consequently reduces load bearing on the articular surfaces. Finally, the body's center of gravity is along a vertical line passing anterior to the knee joint. All these factors amount to optimizing the physiological work required by the joint in maintaining an erect posture. This is how all the soft tissues including ligaments, menisci and cartilage are involved in the locking function.

2.1.3 Diseases of the knee

Cartilage degeneration

The knee is among the most common sites of osteoarthritis, which is the degeneration of the articular cartilage. Degeneration occurs if there is abnormal force acting on a normal, healthy joint or a normal force on an abnormal cartilage. Although the etiology of the osteoarthritis disease is not clear, common associations include age related cartilage changes, overuse or underuse of joints, pre-existing articular deformity, joint trauma and genetic factors.

Soft tissue injury

Any sort of tissue disruption around the knee can affect the stability of the joint. It can also mark the start of a degenerative disease. Soft tissue injury can also affect the neurovascular bundle and

subsequently influence proprioception. Since the knee is composed of a bundle of soft tissues, it is easily susceptible to any extreme forces around the area which might result in a ligament or meniscal tear.

2.2 Etiology and mechanics of cartilage degeneration

Osteoarthritis is one of the most common diseases associated with the articular cartilage of the knee. Cartilage degeneration has been attributed to various reasons such as age, overuse, occupations, ligament tears, and strenuous activities of daily living. In general, the mechanical environment has an influence on the state of the cartilage (Felson, Lawrence et al. 2000; Brandt, Radin et al. 2006). The etiology of cumulative trauma may be attributed to the viscoelastic nature of joint tissues (Kumar 1990). Articular cartilage of the knee is viscoelastic (Hayes and Mockros 1971) and mechanical loading on the knee influences the degeneration of the cartilage *in vivo*. Activities such as kneeling and squatting have shown a greater propensity for cartilage degeneration than daily activities like walking and standing (Eckstein, Lemberger et al. 2005; Hosseini, Van de Velde et al. 2010). However, to date there has been no evidence of a functional relationship between mechanical loading on the knee and *in vivo* cartilage degeneration. Such a functional relationship could help determine at what stage of loading the cartilage would be susceptible to damage. Knowledge of the functional integrity of the joint *in vivo* may help to determine how the cartilage reacts to loading over time. Previous studies have demonstrated the use of MRI scans for assessing cartilage deformation *in vivo*. These methods however are restricted to laboratory environments and constrained by the need for expensive equipment. Alternate non-destructive testing methods such as vibration emission may be a worthwhile option to explore for *in vivo* joint integrity assessments.

MRI scans and X-rays are used to clinically assess the state of cartilage degeneration but these are expensive methods that require elaborate equipment, and cannot be used *in vivo*. They do not indicate the *in vivo* behavior of the joint while it degenerates. Such functional integrity of a joint *in vivo* is useful to know, in order to gather time-dependent data.

Cartilage degeneration

An integrated perspective of the structural, biochemical and biomechanical pathways involved in degeneration is essential for understanding the *in vivo* response of the cartilage to its mechanical

environments (Andriacchi *et.al.*,2004). Cartilage degeneration can result in a change of soft tissue properties and this can distort joint congruency and affect tolerance of tissues. In order to maintain a healthy cartilage, there must be equilibrium between the degeneration and synthesis of cartilage constituents. The cells of the cartilage, called chondrocytes participate in retaining this equilibrium by synthesizing collagen and proteoglycans which are essential components of the cartilage cellular matrix. According to Andriacchi (year), degeneration can happen in two stages – *initiation* and *progression*. Cartilage adapts to mechanical stimuli and initiation of cartilage degeneration can occur due to abnormal loading on the joint. For instance, due to a soft tissue injury such as ACL tear, load bearing may shift to a part of the cartilage which is not adapted to normal day to day loading. The load threshold of the unadapted cartilage may not be as high as the other parts and can be susceptible to further degeneration. The frequency of load bearing in an area determines variations in cartilage morphology. Initiation can also happen without an obvious or acute mechanical trauma. It could be due to a chronic shift in load bearing to a region which is not well adapted to normal daily loads. Occupational activities which are strenuous and performed in awkward postures, could thus lead to such a chronic shift in load bearing. Progression can be result of cumulative loading over those parts which have already started to degenerate. Degeneration of the cartilage results in loss of proteoglycans and disruption of the cartilage cellular matrix. This increases friction between the cartilage surfaces and subsequently, the tangential shear force increases, causing fibrillation of tissues. This leads to increased compressive forces on decreased tissue tolerances (Andriacchi, Mündermann et al. 2004). As a result any small discrepancy in the cartilage tissues might negatively affect the health of the joint unless protective measures are followed.

One of the reasons for loss of cartilage health could be insufficient rest for the tissues between tasks. Prolonged static loads can lead to accumulated damage without appropriate rest to release the stress in the passive tissues which are posture dependent (Callaghan and McGill 2001). Passive tissues or tissues which do not voluntarily activate such as cartilage, tendons and ligaments make up a considerable part of load-bearing in the knee. Prolonged squatting without appropriate rest can cause accumulated damage to these tissues. Furthermore, a lack of low load bearing rest could ultimately affect the synthesis of proteoglycans, ultimately disturbing the equilibrium between degeneration and repair.

Degeneration in deep flexion

Many studies have found that deep knee flexion bears an association with cartilage degeneration. Researchers have used magnetic resonance imaging (MRI) to examine cartilage morphology following a deep knee flexion task. However, MRI scans have limitations. They are not always accurate representations of *in vivo* deformation and they do not give early information on the onset of disease.

Eckstein and group used MRI images to examine *in vivo* patellofemoral as well as tibiofemoral cartilage deformation in activities involving deep flexion such as squatting, walking, running and exercise (Eckstein, Lemberger et al. 2005). They found a *dose-response* relationship for patellar cartilage deformation. The area of patellar degeneration was larger for higher ranges of flexion. However, the tibiofemoral deformation was found to be significantly less than the patellar cartilage deformation. This could probably be attributed to the adaptation of the tibiofemoral area to activities of daily living. Stress on the patellar area was found to act especially during deep flexion which is not as common as daily activity. As a consequence, this part of the cartilage may be abnormally loaded during deep flexion (Andriacchi, Mündermann et al. 2004). However, Eckstein et al. also report that no significant difference was observed in cartilage deformation properties between physically fit athletes and non-athletes.

Joint moments have been observed to peak during deep knee flexion activities and pathological changes to the cartilage have been attributed to such physical demands (Nagura, Dyrby et al. 2002). The tibiofemoral contact area is reduced by 55% when the knee is flexed to 90 degrees. This could explain the high magnitude of loading pressure on the patella-femoral area and joint pathology as a consequence. The loads during deep flexion are therefore important considerations in the pathogenesis of the knees. Seedhom *et al* (2006) propose that sudden bursts of high stress activity can be very damaging because it would apply pressure on an unconditioned cartilage. Secondary movements of the knee such as medial or lateral rotation depend on the angle of flexion, muscle activity and task being performed (Dyrby and Andriacchi 2004). Thus, the state of the cartilage during weight bearing depends on its kinematic and kinetic environment. Cases of knee pain had greater radiographic scores than controls, implying poorer cartilage health; quadriceps strength influences the causation of pain in the knees (O'Reilly, Jones et al. 1998). This could mean that if the load bearing function of the cartilage is affected due to occupational exposures, some of the load could be transferred to the quadriceps resulting

in their fatigue. The more the active muscle tissue fatigues, the more is the amount of load which the soft tissues of the knee would have to withstand.

2.3 Cumulative exposure and cartilage degeneration

2.3.1 Cumulative exposure

Cumulative loading on the knee in kneeling/squatting has been quantified by various methods in the past. Questionnaires and interviews have been used to record task details including frequency of lifts, weight of lifted/carried objects and duration of lift (Seidler, Bolm-Audorff et al. 2008; Klusmann, Gebhardt et al. 2010). A cumulative exposure estimate is made using these details. However, these are self-reported data and have some limitations. The accuracy and applicability of such data have been questioned for various reasons such as questionnaire reliability (Van der Beek and Frings-Dresen, 1998). The cases might overestimate their exposure because of increased psychophysical or physiological demands. Controls might underestimate details such as duration of work. This is called recall bias. Questionnaires and interviews also depend upon human memory and therefore, variations are bound to occur within a homogeneous population group. Videotaping has been used to elicit details for computing cumulative exposure. Although such a method eliminates recall bias and provides relatively objective data, the exposure measurement obtained might be just sufficient to establish an association with disease. The measure would not be suitable for accurately assessing the degree of exposure to which the joint has been subject to. This is because the actual measure of exposure can depend upon biomechanical factors such as joint angles and external joint moments induced by the specific task. The knee adduction moment has been used as a proxy for estimating cumulative load on the knee while walking (Robbins, Birmingham et al. 2009). However, an equivalent proxy moment can be hard to find in case of uncommon postures owing to the numerous muscle groups involved in creating the posture.

In order to estimate the compressive or shear stress on joint tissues such as the low back, biomechanical models have been developed in the past. Although there have been models developed to predict both acute as well as cumulative loading on joints, not much is known yet about the mechanical tolerance of tissues under repeated loading (Chaffin 2006).

Biological tissues are viscoelastic in nature (Kumar 1990). This means that when the body's soft tissues are subjected to any kind of load repetitively or for a prolonged period of time, the time history of the stresses results in some residual deformation in the tissues. Ligaments can slacken and hydration of tissues can get affected because of prolonged loads (Kumar 1990). This decreases the mechanical threshold of tissue failure. In this way, even moderate stresses when exerted over a long period of time can decrease the threshold of tissue failure. This is called cumulative loading and it can explain why a routine activity may suddenly cause pain or injury after several years of doing it (Kumar, 1990).

Cumulative trauma is caused due to forceful, repetitive or prolonged motions. Repetitive exposure would lead to micro trauma whereas prolonged exposure is associated with excess strain and creep of tissues (Norman, Wells et al. 1998). It is more often a result of combined loading (Kroemer 1989). Studies have found epidemiological evidence which shows that combined lifting and squatting results in reduced strength and psychophysical lifting capacity as compared to lifting or squatting alone (Gallagher 2005). Cumulative trauma can result in compression of nerves and impede blood vessels. It can manifest as joint inflammation, joint pain and muscle pain. As a result, a person with such trauma has difficulty performing even simple day-to-day activities.

Occupational activities such as lifting, stooping, squatting, kneeling, pushing, pulling and carrying are biomechanical factors which can increase the propensity of tissue failure in joints over time. Therefore, there has been a lot of interest in quantifying cumulative joint exposures in occupations.

The knee is a joint which performs the function of load bearing and is mainly supported by soft tissues including cartilage, ligaments, and menisci. All of these are viscoelastic in nature. When there is frequent stress over a period of several months or years, as in the case of occupational loads, the mechanical tolerance of these tissues is decreased. Lifting activities, squatting and kneeling postures activities are known to exert high torques on the knee (Tak, Paquet et al. 2009). This high external moment on the knee would correspond to high internal compression or shear forces on the load bearing surfaces of the knee.

Several occupations such as agriculture, construction and mining involve prolonged squatting activity. These occupations have been associated with a higher risk of osteoarthritis than others. Osteoarthritis is a chronic musculoskeletal disease which manifests as wear of the articular

cartilage, often resulting in joint pain and is associated with several biomechanical causal factors. Moreover, squatting or any sort of deep flexion activity has been associated with increased degeneration of the cartilage (Nagura, Dyrby et al. 2002; Bingham, Pappanagari et al. 2008). Therefore, cumulative loading exposure in such occupations can be of interest for understanding the etiology of cartilage degeneration. Loads acting upon the knee could be several times the body weight. No action limits have been developed for the knee yet but the importance of doing so has been signified (Callaghan, Salewytch et al. 2001). Action limits should consider cumulative load as an integral influential factor in future studies.

2.3.2 Measuring cumulative exposure on joints

Kumar (1990) pioneered in quantifying cumulative exposure on a joint. He calculated low back exposure as a function of compression/shear stress, frequency, weight and time of loading. The compression/ shear stress was calculated by a 2-D mathematical model which was formulated after making cases and controls recall the type of postures they used in occupations. It was found that higher cumulative load ratings correlated with a higher incidence of low back pain among the participants. Compression and shear accumulate over time. Following this study, several other researchers have conducted studies to quantify cumulative exposure with a goal to relate its biomechanical causal factors with epidemiological evidence. Some of these works are discussed below.

Norman *et al.* (1998) quantified various biomechanical variables, including peak spinal loads, cumulative spinal loads and kinematic torso variables of exposure, for auto assembly workers. They used a time integral of moments for calculating the cumulative load. One of their main findings was that cumulative load seemed a distinct measure from the rest in distinguishing cases of low back pain from controls. Another important finding was that there is no distinct preference in measurement between kinematic variables such as hand force, and internal compression or shear forces experienced by the tissue. Both external and internal types of measures had their own impact on information about exposure. However, the effects of repetitive and prolonged exposures could not be separated.

Recently there has been interest in quantifying cumulative loads on the knee. Cumulative exposure to daily activity has been quantified in terms of walking time and number of steps per day (Robbins, Birmingham et al. 2009). Their estimate, however, was based on the knee adduction moment which is not the appropriate proxy for loading under constrained postures. Exposure for tasks performed in such postures has been usually classified in terms of job titles or self-reports of work requirements. However, these methods do not provide a quantitative estimate for exposure-response relationships in terms of loading intensity or duration of exposure. Although it is known that biomechanical variables such as the torque on the knee increase for manual material handling activities, little is known about their relation to epidemiological risk associated with activities in unusual postures (Tak, Paquet et al. 2009) . In other words, nothing is known about the limits of exposure for a particular operation while straining the knees. Moreover, obtaining quantitative measures can be time consuming because of the laboratory nature of experiments.

Tak *et al.* (2009) analyzed possible sources of variability in knee pain among construction workers. They quantified exposure in terms of the percentage of time spent in kneeling/squatting/combined positions with respect to operations, tasks, workers and trade. They found that the greatest variance in knee pain existed for the task type and operation performed. This is relevant because it informs us that variables like task type are important considerations for estimating cumulative exposure as a consequence of their variability properties (Tak, Paquet et al. 2009).

Variance exists among methods of estimating cumulative load and these can influence the reliability of a dose response relationship between cumulative exposure and trauma. While quantifying cumulative loads, possible variance in the methods used should be accounted for. Reducing the sampling rate at which data is collected has shown to improve the accuracy of exposure measurement. (Callaghan, Salewytch et al. 2001). Despite all the attempts at quantifying cumulative loads on joints, to date no study has successfully developed a functional relationship representing the effect of joint loading on internal pathology such as pain or tissue tolerance. This is because it has been difficult so far to monitor time sensitive responses of the body to joint loading. The subjective nature of data sources such as job titles or self-reports, and

the time consuming nature of formal experimentation are contributors to this difficulty (Van der Beek and Frings-Dresen, 1998).

2.4 Introduction to non-destructive testing methods and vibration emission

Non-destructive testing (NDT) methods are scientific methods used to test and evaluate material properties without causing any damage to the material. Examples of NDT methods include ultrasonic testing, magnetic-particle testing, eddy current testing, liquid penetrant testing and radiographic tests.

The non-invasive nature of these tests makes them very useful for detecting damage and wear in machines and equipment without causing any damage. Acoustic emission (AE) is one such popular method of in vivo monitoring of equipment. It is the emission of elastic stress waves due to release of strain energy during deformation, phase change or crack formation (Everson and Hoessein Cheraghi 1999).

Vibration emission has been used for assessing degradation in gearboxes (Loutas, Roulias et al. 2010). The vibration emission parameters included combinations of statistical moments of signal series and their Fourier transforms. Features were extracted on the basis of the monotonicity of parameters and those parameters which correlated with gear box wear were chosen.

AE is becoming increasingly common in automated tool wear monitoring. In a real time precision drilling process (Cheraghi and Everson 1999), the diameter of the hole drilled in steel (i.e. quality of hole drilled by the robot) is correlated with an acoustic emission signal parameter. Simple parameters such as counts, hits above a threshold, RMS and energy of the wave were used for measuring acoustic emission.

In the clinical field, AE can be useful for distinguishing healthy knees from diseased ones. (Franke, Dorner et al. 2004; Mascaro, Prior et al. 2009). Franke *et al* (2004) conducted a series of acoustic emission studies upon placing a transducer on the knees for detecting cartilage lesions and effect of loading on the knees. Thus, there is some evidence that mechanical activity can cause changes in the articular cartilage, which are detectable in a healthy knee. For the purpose

of diagnosis, the energy and frequency of the emission spectrum are representative enough of cartilage properties (Reddy et al., 1995; McCoy et al., 1987).

2.5 Vibration emission in joints

2.5.1 Physics behind vibration emissions.

Vibration emission is a NDT method commonly used for monitoring the health of industrial machines which are frequently subject to dynamic loads over long periods of time. It is the release of strain energy from a surface which has been subjected to external stress increasing the roughness of the surface. Friction between two surfaces produces elastic waves which can contain information about the state of wear or cracks present in the surfaces. In the case of the knee, the articular cartilage functions as a major component of dynamic load bearing. In a diseased knee, the cartilage morphology changes at a microscopic level and the lubrication of tissues is affected (Felson, Lawrence et al. 2000). Therefore, the friction between the cartilage surfaces increases and so does the progression of cartilage wear. Elastic waves of a wide range of frequencies are released due to daily movement of the abnormal friction surfaces. Eventually, the thickness of the cartilage decreases (Mascaro, Prior et al. 2009). Vibration emissions contain information on the state of lubrication and joint integrity of the cartilage. Moreover, the knee has been said to be the most accessible part and most susceptible to joint disease (Blodgett 1902).

2.5.2 History of joint vibration emission

In condition monitoring of machinery, researchers in the past have used various methods for extracting features and parameters of acoustic emission (Carpinteri, Lacidogna et al. 2007; Loutas, Roulias et al. 2010). In the clinical field, accurate diagnosis of the knee to classify degenerative diseases has required invasive technology such as arthroscopy. MRI, X-Ray and radiography, although non-invasive, are expensive and may not give perceivable results until into the degenerative stage. Further, none of these methods give early information of a possible loss of cartilage health. Early information with regard to knee joint degeneration is becoming increasingly important for taking precaution before the stage of degeneration begins. Above all, the rising interest in acoustic signals from the knee joint is because it is hoped that this method would emerge as a non-invasive, pain free technique to evaluate joint sounds objectively and in vivo.

Physicians have attempted to classify abnormal knees from healthy ones based on joint sounds produced due to degeneration of the cartilage and soft tissue components inside the knee. Listening to the sounds produced by the knee is called joint auscultation and Blodgett was among the first to study knee noises using a stethoscope. Various other researchers in the earlier half of the 20th century examined the same. However, stethoscope technology was not enough to filter out noises from skin friction, snapping tendons, muscle vibrations and hand tremors (McCoy, McCrea et al. 1987).

Chu *et al.* (1978) were probably the first group who attempted to quantify cartilage roughness in the form of electric signals. This was a more objective technique to quantify the sounds than the use of stethoscopes. They used microphones in the audible frequency range upon the patient's patellar region to detect knee sounds. By using the audible frequency range, the noise of snapping tendons which is said to be of low frequency was avoided. Degenerative knees showed a higher power than rheumatoid and normal knees.

Accelerometers have been used in the past to capture knee sounds since these instruments were capable of capturing low frequency vibrations which are an integral component of knee joint sounds (McCoy, McCrea et al. 1987; Kernohan, Barr et al. 1991). The results of McCoy *et al.* (1987) showed that the nature of these knee signals matched with the corresponding physiological conditions of normal and pathological knees. They concluded that the "patellofemoral crepitus" is an important signal which contains information on the state of degeneration of the cartilage and that its quality, amplitude and distribution reflect cartilage integrity. With deterioration, the amplitude and frequency of the crepitus was found to decrease. Some studies have assessed the integrity of joint implants. Keeping track of the health of the implant is analogous to condition monitoring of industrial bearings and machine structures. A vibrating probe with certain input frequencies was used in combination with an accelerometer for output, to assess joint integrity of implants (Georgiou and Cunningham 2001). Intact implants showed just a fundamental frequency while loosening implants resulted in more than one harmonic. Another group (Glaser, Komistek et al. 2010) evaluated hip joint conditions using

accelerometers on bony prominences. They also collected gait data and found matches between extreme instances during the gait cycle and high frequency vibration signals.

In the last two decades, researchers have explored and evolved possibilities of quantifying low frequency vibration signals from the knee (Tavathia, Rangayyan et al. 1992; Zhang, Frank et al. 1992; Grodzinsky, Levenston et al. 2000; Rangayyan and Wu 2008). These methods have used accelerometers to quantify knee joint vibrations.

3.0 METHODS

3.1 Participants

The study protocols were approved by the Institutional Review Board, Virginia Tech (Appendix B). Two volunteers were recruited for the pilot study. Although the pilot participants were assigned with eight tasks, one of the participants quit after 4 tasks and another after 6 tasks. After the pilot study, flyers were posted around the Virginia Tech campus and a call for participants was sent via the graduate school list-serve. There was no constraint on gender, as long as the participant was deemed physically and mentally fit for performing the repetitive squatting tasks. Twelve college males were recruited for the final study, based on the results of a demographic questionnaire (Appendix A) answered by interested individuals. To isolate the effect of loading on the joint from any other pathological conditions, the inclusion criteria was to recruit only those individuals aged 18-30 years, who exercised regularly for at least 2-3 times a week, and did not have any reported history of musculoskeletal trauma. Participants were reportedly free of any other medical conditions. Effects of recent exercise and diurnal changes on the knee joint condition were not considered in this experiment. Mean (SD) age, height and body mass of participants was 24.3 (2.0) years, 175.2 (7.9) cm and 73.4 (6.9) kg. Of the twelve participants in the final study, one participant's data was discarded due to erroneous measurements. Of the remaining eleven participants, ten participants performed up to 4 tasks, while eight participants performed up to 5 tasks and only three participants performed all the six tasks.

3.2 Instrumentation

3.2.1 Motion capture system

Passive Marker System

Kinematic data from all lifting tasks were captured using a passive marker seven-camera motion capture system (Vicon MX-GIGANET, Vicon Motion Systems, Los Angeles, CA, USA).

Reflective markers were affixed on the following anatomical positions of the right leg as per Dumas et al. (2007): Calcaneus, Toe tip, Lateral Malleolus, Sphyrion, Lateral Epicondyle, Medial Epicondyle and a cluster of markers as a triangle was used for the shank.

Force Plate

Ground reaction force data were collected while the participant performed the assigned tasks standing on two force platforms (AMTI MSA-6, Watertown, MA, USA). Kinetic data sampled at a rate of 1000Hz, were collected while the participant performed the task with one foot on each plate. Right leg data was used for analysis. The analog signals were converted to digital form by a data acquisition card (Vicon Motion Systems, Los Angeles, CA, USA) and the data collection was managed by Vicon-Nexus (Version 1.6.1, Vicon Motion Systems, Los Angeles, CA, USA).

3.2.2 Accelerometer

An ICP or Integrated Circuit Piezoelectric is an accelerometer class wherein the electronic amplifier circuitry is built into the accelerometer. ICP accelerometers are unique for their use of a single coaxial cable for power input and signal output which returns several advantages such as small size, light weight and low cost cabling. The piezoelectric device used (Figure 3.1) was a miniature, lightweight (0.8gm) ICP accelerometer. ICP standard sensors come with built-in circuitry for amplification or signal conditioning, hence eliminating the need for building an external unit. The device (PCB 352-A24, PCB Piezotronics Inc., Depew, NY, USA) had an input amplitude range of $\pm 50g$ and a bandwidth of 1-8000Hz with a sensitivity of 100mV/g. These specifications were similar to those devices used previously for differentiating between pathological and normal knee joints (Nieminen, Saarakkala et al. 2004). In order to ensure participant safety, the accelerometer chosen was case-isolated with a layer of non-conducting ceramic.

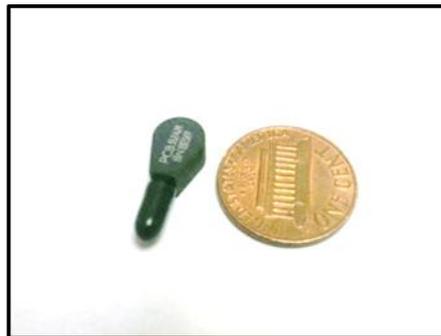


Figure 3.1: Accelerometer unit used in the study

The accelerometer was powered by a 4-channel line power unit (PCB 482-C05, PCB Piezotronics, Depew, NY, USA). Joint vibration signals were collected at a sampling rate of 5000Hz through a custom-built oscilloscope program in LABVIEW (Version 8.5, National Instruments Inc., Austin, TX, USA). The signals were filtered using a 2nd order Butterworth filter and subsequently processed in MATLAB (MATLAB R2010a, MathWorks Inc., Natick, MA, USA).

3.2.3 Metronome

Two types of metronomes were used in the study.

1. A virtual metronome set to 20 lifts/minute and 40lifts/minute was used to train and enable participants to perform tasks at high and low lifting frequencies.
2. A digital metronome set to 60 beats/minute was used to train and enable the participants to swing their leg at a consistent speed while recording signals from the accelerometer.

3.3 Pilot study

3.3.1 Procedure

Capturing vibration signals

Two participants were recruited for the pilot study, based on the inclusion criteria discussed earlier. After informed consent, the participant was seated on a raised seat such that the feet were suspended freely at rest. Baseline joint vibration signals were recorded from the participant's right knee at the mid-patella and medial condyle. All joint vibration signals were recorded in a reference movement performed by the participant. This involved swinging the right leg through one cycle of knee extension and flexion over a 90⁰ range of motion at a consistent speed of 60 beats per minute. Three such trials of the vibration signal from the patella and medial condyle were recorded after each task was performed, to record the repeatability of measurements.

Inducing cumulative mechanical load

Each participant performed a series of dynamic squat-lifting tasks with box-weights. Each task was a combination of a certain box weight and frequency of lift. Participants were assigned each task for a period of eight minutes, after which vibration signals were recorded. Participants were instructed to follow the metronome and keep their back as straight as possible; to enable them to

use their knees as much as possible during the lift. However, of the eight tasks assigned, one of the participants performed only four tasks while the other performed eight tasks.

Choice of mechanical loading

There were four box-weights and two lifting frequencies and the eight possible combinations of tasks as summarized in Table 3.1. The box weights and lifting frequencies were chosen so as to ensure that the product of load and frequency are a spread of values over a wide range rather than being concentrated in a narrow band of cumulative exposure. This was performed in an effort to ensure multiple levels of cumulative load as defined by Kumar (1990).

Table 3.1: Tasks assigned based on combinations of load and frequency of lifts.

	Box + No Weight	Box + 1 Weight	Box + 2 Weights	Box + 3 Weights
Low Frequency	1.5kg x 20lifts/min	4.5kg x 20lifts/min	7.5kg x 20lifts/min	10.5kg x 20lifts/min
High Frequency	1.5kg x 40lifts/min	4.5kg x 40lifts/min	7.5kg x 40lifts/min	10.5kg x 40lifts/min

3.3.2 Dependent variables and predictor variables

The dependent variables considered in the pilot study included 1) The relative Root Mean Square (relative RMS) power of the signal 2) Relative Median Frequency (RMF) of the power spectrum of the signal. These are computed as described below.

Relative Root Mean Square (RMS) power

The instantaneous power of the vibration signal is the square of its amplitude after filtering the signal. The mean power of the signal was calculated as the average of the instantaneous powers in time series. The RMS power for each trial was computed as the square root of the mean power of the signal. The average RMS power of the three trials captured after each task was recorded for analysis.

Relative Root Mean Square (RMS) Power

$$= \frac{\text{RMS of the vibration signal recorded after a task}}{\text{RMS of the vibration signal recorded at the start of the experiment}}$$

Relative Median Frequency

The Fast Fourier Transform of the joint vibration signal in time series returns the frequency spectrum of the signal. The median frequency of the signal is defined as that frequency which separates the total power of this frequency spectrum into half. The relative median frequency of the 'n'th task was defined as the ratio of the median frequency after 'n' tasks and the initial median frequency at the beginning before performing any of the tasks.

Relative Median Frequency

$$= \frac{\text{Median Frequency of the vibration signal recorded after a task}}{\text{Median Frequency of the vibration signal recorded at the start of the experiment}}$$

The predictor variable set for the study was the 3-D cumulative moment on the knee which was computed as the integral of the 3-D instantaneous moment time series.

3.3.3 Data analyses

Joint Vibrations

The vibration signal captured was filtered with a 2nd order Butterworth band pass filter in the range 20Hz to 1000Hz. The RMS power was computed for the entire trial of the signal in time series and the median frequency was obtained from the power spectrum.

3.3.4 Pilot results

In the time domain of the signal, the power of each trial (one complete extension and flexion) was computed in terms of RMS and the in the frequency domain, median frequency of each trial was calculated. The means and standard deviations of the Relative Median Frequency (RMF) for all subjects are plotted against the corresponding loading levels. The RMS and RMF plots can be observed in figures 4.1 (a) and (b). There is a decline in the RMF with progressing levels of loading, whereas the RMS does not display any such trend in data. Therefore, the RMS was discarded and only the RMF was used for further investigation.

Table 3.2 (a): Sample RMS values obtained for the patella.

Task	0 (6 min)	1 (12 min)	2 (18 min)	3 (24 min)	4 (30 min)
Trial 1	0.15 V	0.22 V	0.25 V	0.26 V	0.23 V
Trial 2	0.13 V	0.17 V	0.22 V	0.24 V	0.25 V
Trial 3	0.12 V	0.18 V	0.33 V	0.31 V	0.34 V
Average	0.13 V	0.19 V	0.27 V	0.27 V	0.28 V
Std. Dev	0.01 V	0.02 V	0.04 V	0.03 V	0.05 V

Table 3.2 (b): Sample Median Frequency values obtained for the patella.

Task	0 (6 min)	1 (12 min)	2 (18 min)	3 (24 min)	4 (30 min)
Trial 1	89.1696	49.1683	49.5017	52.0017	70.3357
Trial 2	136.3379	61.5021	47.1682	40.3347	83.8361
Trial 3	95.5032	53.6685	42.3347	40.8347	91.5031
Average	107.0036	54.77963	46.33487	44.39037	81.89163
Std. Dev	25.6009	6.241524	3.655449	6.596347	10.71683

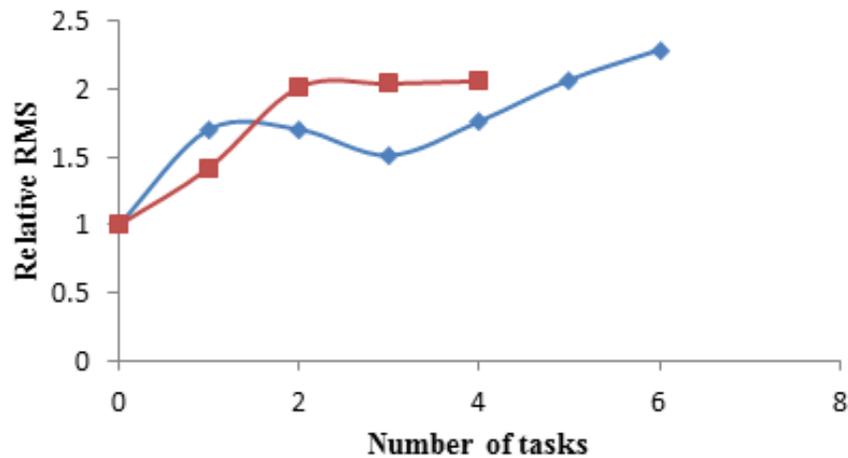


Figure 3.2 (a): Relative root mean square (RMS) power of the signals from the patella measured after each task. The RMS power here is normalized to its initial value computed from the signals which were recorded at the start of the experiment, before the participant performed any tasks.

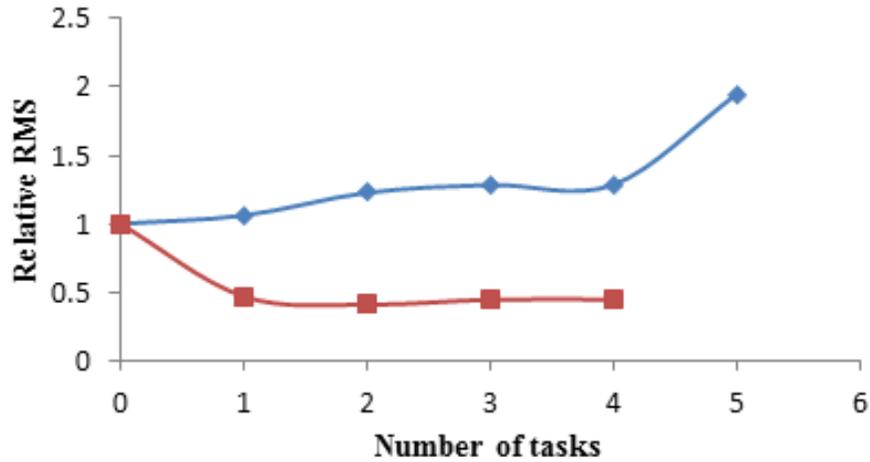


Figure 3.2 (b): Relative root mean square RMS power of the signals from the medial condyle measured after each task. The RMS power here is normalized to its initial value computed from the signals which were recorded at the start of the experiment, before the participant performed any tasks.

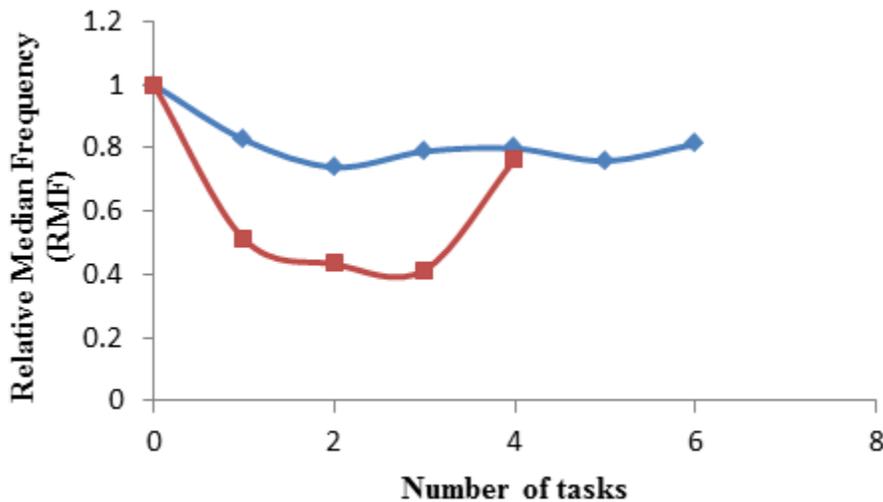


Figure 3.2(c): Relative Median Frequency (RMF) of signals measured from the medial condyle after each task. The median frequency is normalized with respect to its initial value which is computed from the signals recorded at the start of the experiment, before the participant performs any tasks.

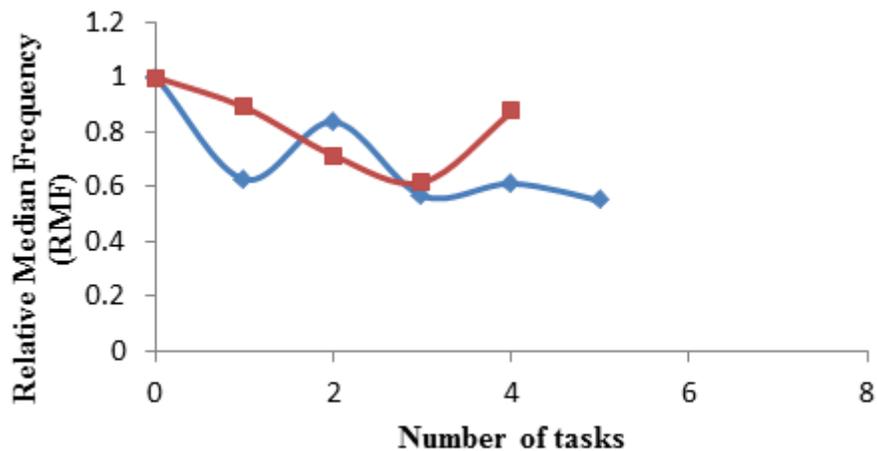


Figure 3.2(d): Relative Median Frequency (RMF) of signals measured from the patella after each task. The median frequency is normalized with respect to its initial value which is computed from the signals recorded at the start of the experiment, before the participant performs any tasks.

3.3.5 Issues with pilot and revisions made

First, the box-weights were replaced with free weights since lifting the boxes became unwieldy and participants tended to deep squat with their thighs impinging upon their legs.

Secondly, the task duration was reduced to six minutes from eight since the participants found it inconvenient to perform a stretch of eight straight minutes.

Thirdly, only the Median Frequency was considered for further investigation since the RMS values did not show a consistent trend across signals captured from the patella and medial condyle.

3.4 Final experimental procedure

Briefly speaking, each participant had to perform a series of squat-lifting tasks on a force platform; kinematic and kinetic data were collected continuously throughout the task. Vibration signals were then recorded immediately following each task.

Calibration

To maximize the accuracy of data collection, the laboratory space used for the lifting tasks was calibrated before the participant arrived. This included static and dynamic calibration using the 5 marker wand system, and software calibration algorithms provided by the manufacturer.

Marker Placement

After informed consent was obtained from the participants, reflective markers were placed on anatomical positions of the foot and shank for each leg (as recommended in Dumas et al., 2007)

Placement of accelerometers

The accelerometer was affixed to the mid-patella and medial condyle positions as shown in Figures 3.2 (a) and (b) using double sided tape. Petro wax was used to mount the accelerometer firmly onto the tape. The positions were marked before start of the experiment so as to maintain consistency in collecting data from the same position after every task.

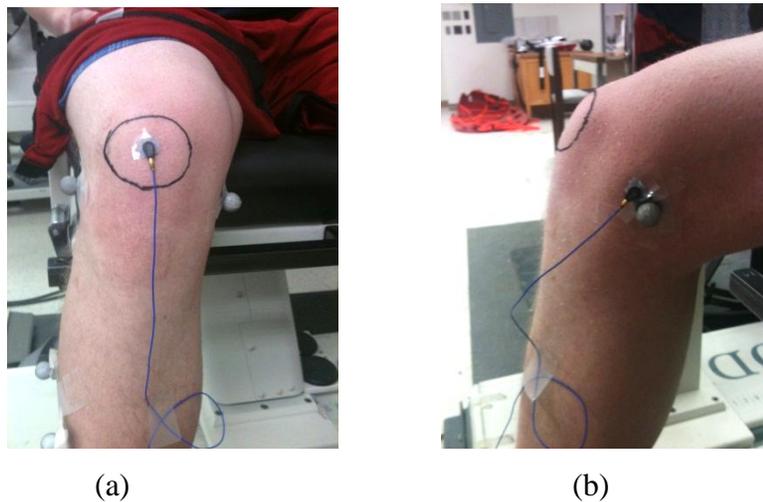


Figure 3.3: (a) Accelerometer positioned on the mid-patella. (b) Accelerometer positioned on the medial condyle.

The accelerometer positions were chosen such that there was a bony prominence with little underlying fat in the area. Since vibrations from various parts of the joint ultimately transmit to the patella, (Rangayyan et al., 1992) the mid patella was marked and utilized for measurements.

Reference movement

All acoustic signals were collected while the participant performed a reference movement. Each participant was instructed to swing the right leg for one movement cycle with a constant speed of 60beats/min, following the cadence given by a metronome. One movement cycle included one flexion-extension movement through an approximately 90⁰ range of motion. Each cycle lasted for about six seconds (i.e., 3 metronome beats up, 3 metronome beats down). Participants were instructed to keep the movement as smooth as possible without any jerking motion. Every participant was allowed to practice the movement several times until they were able to swing their leg smoothly and with a consistent speed. A consistent speed was ensured by the investigator reinforcing the cadence count verbally when required.

Task Specifications

In order to induce cumulative loading on the joint, task combinations of different loading and lifting speeds were used. The weight of the most commonly lifted cinderblock in US underground coal mines is about 11.3 kg (Gallagher, Pollard et al. 2011). Therefore, the experiment used three comparable weights of 4lbs, 10lbs and 20lbs. A lifting frequency of 20 lifts/min has been used for assessing thigh and trunk range of motion while squatting (Hagen, Sørhagen et al. 1995). Since the experiment attempted to simulate similar restrictive postures, comparable frequency levels of 20 lifts/min and 40 lifts/min were chosen. The frequencies were adjusted in such a way that lifting and lowering could be performed at a comfortable pace. Secondly, the lower frequency of 20 lifts/min attempted to induce static endurance loading on the joint in the squat position while the higher frequency was more dynamic. Both the loads however were cyclic, since they involved lifting and lowering.

The six combinations of dynamic lifts and their cumulative exposure values are represented in the Table 3.1. Cumulative values, representative of external physical exposure were calculated by multiplying the load with the frequency of lift.

Thus, there are in total six loading conditions – combination of three weights and two lifting speeds. The total duration of tasks was therefore effectively around 36 minutes. The duration was set in accordance with literature which suggested that the minimum time required for

viscoelasticity of the human cartilage to stabilize was around 30 minutes (Hayes and Mockros, 1971).

Task Procedure

After recording baseline vibration signals, the participant was taken to the experimental set-up area for squat-lifting exercises as shown in figures 3.3 (a) and (b). The squat lifting procedure was demonstrated to the participants by the investigator with the metronome turned on.

Participants were instructed to place one foot on each force plate while lifting and lowering the free weight pairs of 2lb, 5lb and 10 lb. placed on the table. The position of feet on the force plate was fixed and marked by tape so that all participants would maintain a controlled posture. The table was adjusted to the participant's waist height and boxes positioned below the table fixed the squat height in such a manner that the participants could only perform half-squat. In order to avoid impingement of the thighs on the shank, the squat height was fixed by positioning boxes below the table. Participants were instructed not to bend their back while squat-lifting, so as to allow the knees to bear all the physical demands while minimizing load on the lower back.

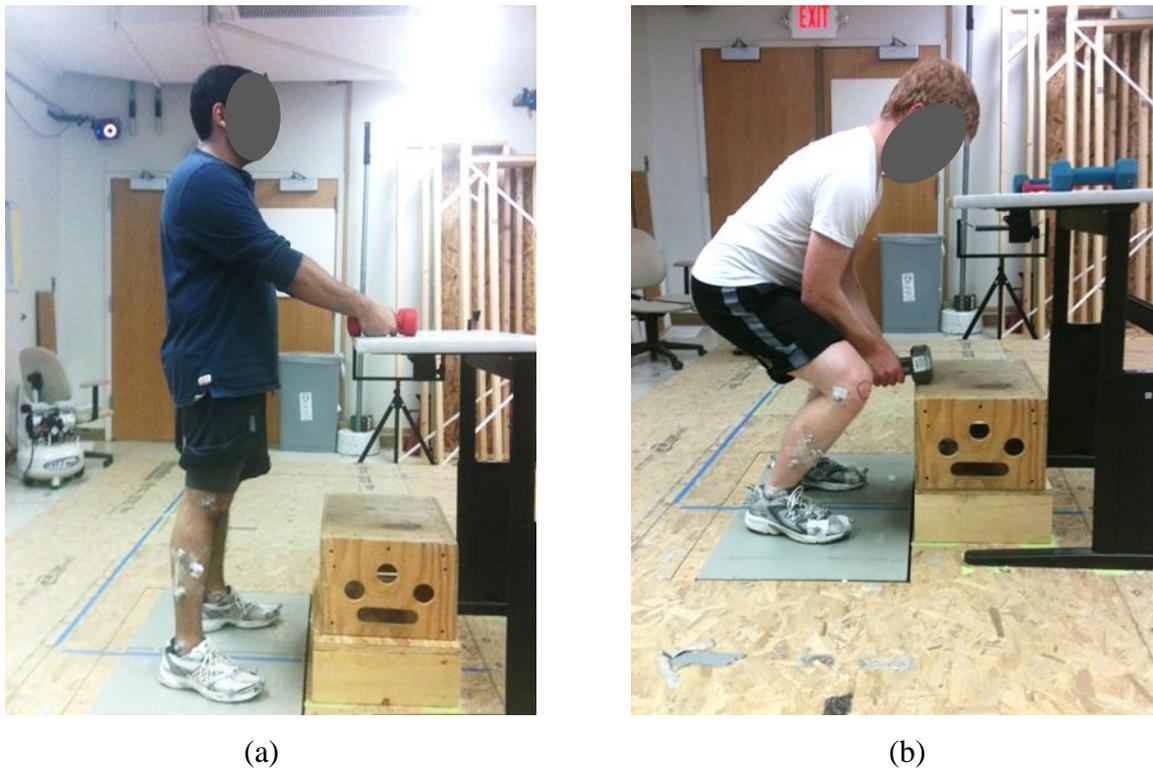


Figure 3.4: Experimental set-up for performing squat-lifting tasks in (a) Resting position and (b) Half-Squat position.

Participants performed each task for six minutes. However, no participant could perform six minutes at a stretch for the 10lb weights and not all could perform 5lbs for six continuous minutes. Therefore, those participants were allowed a few seconds of breathing recovery after every two minutes of loading for the heavier weights. After every 6 minutes of exercise, joint vibration measurements were recorded in the reference movement. Of the total twelve participants recruited, two participants performed 4 out of the six tasks. One participant performed just two tasks. One participant data was discarded due to incorrect measurements.

Table 3.3: Counterbalancing scheme and description of tasks assigned to each individual.

Tasks	1	2	3	4	5	6	Description of task
Participant							
p1, p7	A	B	C	D	E	F	A: 4lbx20lifts/min B: 10lbx20lifts/min C: 20lbx20lifts/min D: 4lbx40lifts/min E: 10lbx 40lifts/min F: 20lbx 40lifts/min
p2,p8	B	F	D	C	A	E	
p3,p9	C	D	E	F	B	A	
p4,p10	D	A	F	E	C	B	
p5,p11	E	C	A	B	F	D	
p6,p12	F	E	B	A	D	C	

3.5 Data analysis

3.5.1 Joint moment calculation (mechanical loading)

The 3-D net external moment at the knee joint was obtained by inverse dynamics using the link segment model (Winter, 2005). The kinematic and kinetic data were sampled by the Vicon-Nexus system software at sampling rates of 100Hz and 1000Hz respectively. Data were filtered with a 2nd order forward and backward Butterworth filter. The kinematic data were low pass filtered with a cut-off frequency of 6Hz while kinetic data were low pass filtered with a cut off frequency of 10Hz. The cut-off frequencies were concurrent with the kinematic and force plate calculations made from similar instrumentation by Preuss and Fung (2004). The kinetic data were then down sampled to 100Hz for ease of matrix computations in MATLAB.

Inverse Dynamics Computations

The filtered kinematic and kinetic data were used as inputs for inverse dynamics computations performed in MATLAB. The sequence of steps for calculating the cumulative 3-D joint moment is described below. The segment masses, segment moment of inertia, and the position of the segment center of mass were determined from scaling equations as per Dumas et al (2007).

Step 1: Calibration and rotation matrices.

Since the passive markers were often blocked from the cameras by the set-up, markers on the medial side of the knee were barely visible in the trajectory. To overcome this issue, a triangle cluster of markers was placed on the shank segment. A single frame from the static trial, when the participant was absolutely steady, was chosen to calibrate the reference matrix. This matrix was used to calibrate the position of knee in space during the dynamic trials.

All the segment lengths and center of mass positions were first determined in the local co-ordinate system as described by Dumas et al (2007). A rotation matrix was computed and applied to obtain these co-ordinates in the global co-ordinate system.

Step 2: Centre of Pressure (COP)

Filtered force data was down-sampled to 100Hz. The center of pressure co-ordinates (x,y,z) in the global co-ordinate system were calculated using the following formulas as prescribed by the AMTI instruction manual for the force plate.

$$COP_x = \frac{M_y}{F_z}$$

$$COP_y = \frac{M_x}{F_z}$$

The thickness of the force plate was assumed to be zero and therefore,

$$COP_z = 0$$

Step 4: Linear and angular accelerations

The angular motion of the lower extremity was resolved into three planes – transverse, sagittal and coronal corresponding to x-y, y-z and z-x planes. Finite difference equations were applied on the position (kinematic) data to obtain the corresponding instantaneous velocities and

accelerations (Winter, 2005). Linear velocities and accelerations of the rigid body segment were computed in the three anatomical planes and used in the final force and moment equations to compute the knee joint moment.

Step 5: Free Body Diagrams

Free body diagrams were drawn in the three anatomical planes to define the forces and moments directly acting upon the foot and the shank. The free body diagrams in the sagittal plane are shown in Figure 3.5.

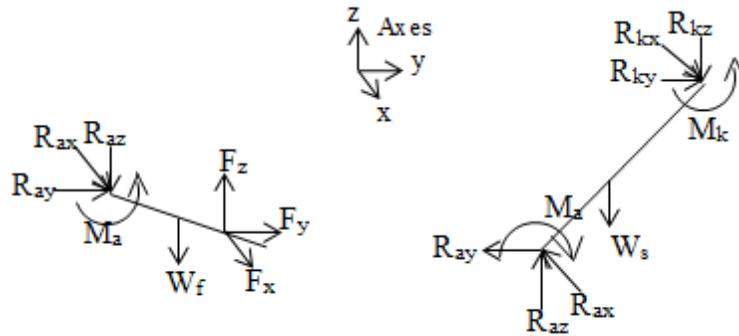


Figure 3.5: Free body diagram showing forces and moments acting on the (a) Foot segment and (b) Shank segment

Step 6: Moment and Force Equations

Based on the free body diagrams, the segment equations are described below.

Foot Segment

Horizontal force equations:

$$R_{ay} + F_y = m_f a_{fy} \text{----- (1)}$$

$$R_{ax} + F_x = m_f a_{fx} \text{----- (2)}$$

Vertical Force equations:

$$F_z - R_{az} - W_f = m_f a_{fz} \text{----- (3)}$$

where m_f is the mass of the foot segment,

a_{fx} , a_{fy} and a_{fz} are the accelerations of the centre of mass of the foot segment in the x, y and z directions respectively.

R_{ax} , R_{ay} , and R_{az} are the reaction forces at the ankle in the x, y and z direction respectively.

F_x , F_y , F_z are the ground reaction forces acting on the foot in the x, y and z direction respectively.

W_f is the weight of the foot segment.

Shank Segment

Horizontal force equations:

$$R_{ky} - R_{ay} = m_s a_{sy} \text{-----} (4)$$

$$R_{kx} - R_{ax} = m_s a_{sx} \text{-----} (5)$$

Vertical force equations:

$$R_{az} - R_{kz} - W_s = m_s a_{sz} \text{-----} (6)$$

where m_s is the mass of the shank segment, a_{sx} , a_{sy} and a_{sz} are the accelerations of the centre of mass of the shank segment in the x, y and z directions respectively.

R_{kx} , R_{ky} , and R_{kz} are the reaction forces at the knee in the x, y and z direction respectively.

R_{ax} , R_{ay} , and R_{az} are the reaction forces at the ankle in the x, y and z direction respectively.

W_s is the weight of the shank segment.

Step 7: Moment Equations

The moment equations are represented in their 3-D vector form, as cross products shown below.

$$\mathbf{M}_a + \mathbf{r1} \times \mathbf{R}_a + \mathbf{r2} \times \mathbf{F} = \mathbf{I}_f \times \boldsymbol{\alpha}_f \text{-----} (7)$$

$$\mathbf{M}_k + \mathbf{s1} \times \mathbf{R}_a + \mathbf{s2} \times \mathbf{R}_k + \mathbf{M}_a = \mathbf{I}_k \times \boldsymbol{\alpha}_s \text{-----} (8)$$

where:

\mathbf{M}_a and \mathbf{M}_k are the 3-D moment vectors at the ankle joint and the knee joint respectively, and these are required unknowns computed using the equations above.

R_a is the vector containing the x, y and z components of the reaction force on the ankle joint.

$$R_a = [R_{ax} \quad R_{ay} \quad R_{az}].$$

Similarly, $F = [F_x \quad F_y \quad F_z]$, $R_k = [R_{kx} \quad R_{ky} \quad R_{kz}]$

I_f and I_k are the inertia tensors about the center of mass of the foot and shank segments respectively. They were obtained from scaling parameters recommended by Dumas et al. (2007).

a_f and a_k are the vectors for segment angular accelerations for the foot and shank respectively.

$r1$ and $r2$ are position vectors denoting the distance of the point of application of R_a and F relative to the center of mass of the foot segment. Similarly, $s1$ and $s2$ are the position vectors for the distances of R_a and R_k relative to the center of mass of the shank segment.

Step 7: 3-D joint moment and cumulative moment

The instantaneous 3-D moment was calculated as the root mean square (RMS) of the moments in each anatomical plane for every sample in the trial.

Thus,

Instantaneous RMS of ankle moments $M_a = \sqrt{M_{ax}^2 + M_{ay}^2 + M_{az}^2}$ ----- (9)

Instantaneous RMS of knee moments $M_k = \sqrt{M_{kx}^2 + M_{ky}^2 + M_{kz}^2}$ ----- (10)

The cumulative moment was the integral of all the instantaneous RMS moments over the time ‘t’ taken to complete the tasks.

Cumulative moment of ankle = $\int_0^t M_a dt$

Cumulative moment of knee = $\int_0^t M_k dt$

3.5.2 Joint vibration signal processing

Data Collection

Before and after the completion of each task, each participant performed the reference movement continuously, following the beat of the metronome. Once the participant attained consistency in

speed and movement, three trials were collected for each signal from the patella and medial condyle during each recording session.

Data Clipping

Each signal was partitioned by visual inspection into extension or flexion, based on ascent or descent of the accelerometer output. Stationary parts at the start and end of the signal which did not have a slope were clipped from the trial. These stationary parts of the signal were from instances during measurement while the leg was resting before changing direction.

Vibration Signal Processing – Based on Ensemble Average

After clipping each signal into flexion and extension parts, the Fast Fourier transform (FFT) between 25 Hz and 2500Hz of each signal was taken. An ensemble average of the FFT at each frequency was computed across ten of the twelve participants for:

- a) Signals recorded at the start of the experiment before any form of loading.
- b) Signals recorded after the participant accomplished 2 tasks.
- b) Signals recorded after the participant accomplished 4 tasks.

For the aforementioned loading conditions, the ensemble average was constructed for signals from the patella, during flexion and extension as well as for signals from the medial condyle during flexion and extension. The purpose of graphing out the ensembles was to identify trends in data across the different loading conditions. Based on the results of the ensemble average, a pre-post t-test and an analysis of variance (ANOVA) of vibration parameters were subsequently performed across the three loading conditions considered for signals from the patella during flexion. The vibration parameters considered for the t-test included the absolute RMS and Median Frequency.

Vibration Signal Processing – Based on different bandwidths

Further, data processing was performed across three bandwidths. First, our pilot results indicated that the median frequencies for readings taken from the patella and medial condyle to be below 100Hz. Since muscle, skin and motion artifacts were to be filtered out, the first bandwidth under consideration was 20-100Hz. Secondly, it has been reported that significant differences between normal and pathological knees occur in the frequency range of 100-500Hz (Reddy et al., 1995).

Therefore the second bandwidth under consideration was 100-500Hz, to test if similar differences could be seen across loading conditions. Thirdly, in an intuitive sense, normal joints would vibrate at higher frequencies than pathological joints. This idea is partially supported by literature which points out that the frequency range of vibrations decreases as the severity of joint pathology increases. Therefore, the third bandwidth under consideration was kept at higher frequencies of 1000-2500Hz.

Similar methods as Reddy et al. (1995) have been used here for processing participant data across three bandwidths. After clipping, data trials comprised of extension and flexion trials. Each raw signal was normalized with respect to its peak amplitude. The normalized signals were filtered in two steps. Firstly, a 4th order Butterworth filter was used to process each data trial of extension and flexion. Data were filtered across three different bandwidths of 20-100Hz, 100-500Hz and 1000-2500Hz. Secondly, a notch filter was designed to eliminate AC power contamination at 60Hz, and harmonics at 120Hz and 180 Hz. After filtering, the frequency spectrum of each trial was obtained by means of the Fourier transform and the median frequency of each signal was recorded. Mean power of each signal was computed as the sum of power spectrum values over each frequency band divided by the total number of frequency components. All these steps are outlined in figure 3.5. Data were processed by a custom program written in MATLAB.

The three bandwidths chosen for this experiment were based on literature and empirical concepts. Chu et al. (1978) used microphones to capture vibrations from the pre-patellar region in one cycle of flexion and extension and suggested that knee joint vibrations exist throughout the audible frequency range of 20-20000Hz and that below 20Hz; the vibrations were primarily due to snapping of tendons during articular movements. Analyses conducted across a wide bandwidth of 20-2500Hz during the pilot and the final study showed median frequencies to exist below 100Hz. Therefore a window of 20-100Hz was used. McCoy et al. (1987) indicated that most pathological vibrations existed below 1000Hz. Reddy et al. (1995) used miniature accelerometers and found significant differences between normal and pathological knee vibrations between 100-500Hz through a simple power spectral analysis. Therefore, this window was chosen to examine if any differences existed across loading conditions.

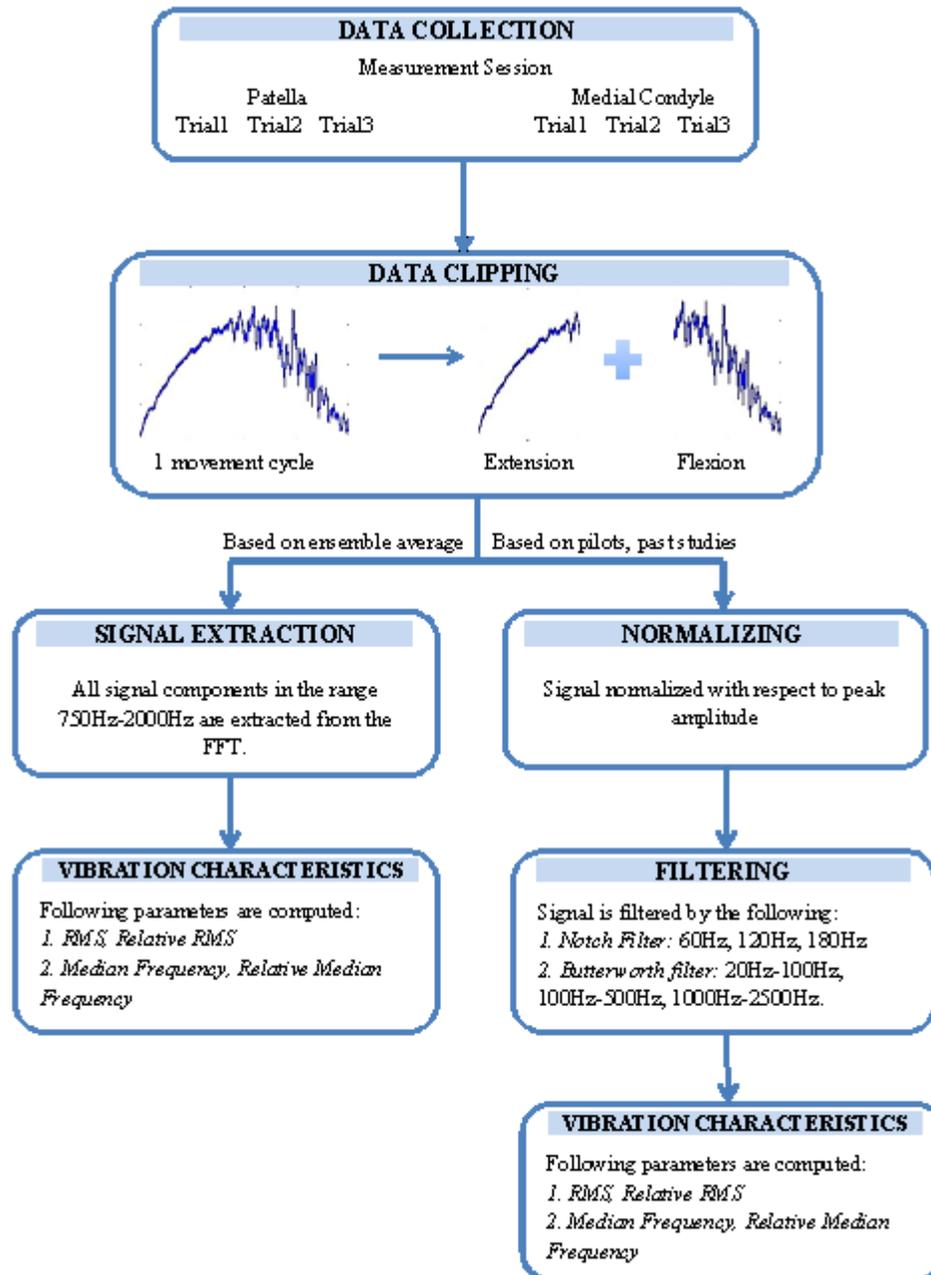


Figure 3.6: Flowchart depicting steps involved in processing vibration data.

3.6 Statistical analysis

The dependent variables under consideration in this study included:

- a) RMS of the vibration signals
- b) Median Frequency of the vibration signals
- c) RMS of the signals normalized to their initial RMS before start of the experiment (Relative RMS)
- d) Median Frequency of the signals normalized to their initial value before start of the experiment (Relative Median Frequency, RMF)

Statistical tests were performed to investigate relationships between cumulative moment and the aforementioned joint vibration parameters. Using data from the ensemble average curves, a two tailed paired t-test was performed comparing the cumulative moment (Nm.s) and variables (a) and (b) before start of the experiment and after 4 tasks for signals recorded during flexion, from the patella in the 750Hz-2000Hz bandwidth. Following this, a one way ANOVA was performed for the variable which showed a significant difference in the t-test. If the ANOVA did not show any differences across means for the non-normalized variables, (a) or (b), the normalized variables were tested by ANOVA. Normalizing the variables with respect to their initial values was an attempt to minimize any carry-over effects of joint deformation.

In order to investigate the possibility of a quantitative relationship between induced moment and vibration parameters, several regression analyses were performed. Linear regressions were performed between the cumulative moment and the relative median frequency (RMF) of joint vibrations across twelve combinations - three bandwidths (20-100Hz, 100-500Hz, 1000-2500Hz), two locations (Patella and Medial Condyle) and two movement types (Extension and Flexion). The regression parameter β was tested for significance in each regression. Coefficients of determination (R^2), root mean square error and standard error of regression were also examined. Residual plots were examined for regression models which exhibited significant associations in order to check for normality in the model. The effect of cumulative load on the vibration parameter, RMF, without the influence of body mass was investigated. In order to do

this, the RMF set of values for each of the regressions which were found significant, were regressed over the cumulative 3-D moment normalized with respect to body mass (Nm.s/kg). Subsequently, a two-factor regression model was performed using the data from the two associations which were found significant. Multicollinearity among the predictor variables was examined for this two factor model.

4.0 RESULTS

4.1 Analysis from ensemble averages

As a part of preliminary investigation of vibration signals, ensemble average of participants' frequency spectra across ten participants for various conditions were computed. The graphs obtained are displayed in Figures 4.1 (a), (b), (c), and (d). The ensemble average plots for signals from the patella, during flexion (Figure 4.1.d.) showed some visually distinguishable differences in magnitude in the 750Hz – 2000Hz range. As a result, signals in this range were extracted and their RMS and median frequency were computed for further analysis.

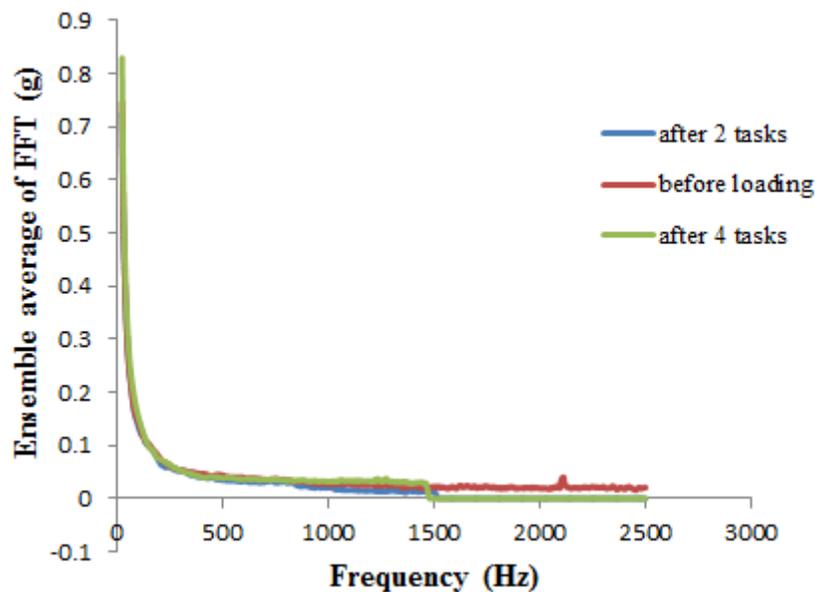


Figure 4.1(a): Ensemble average of the Fourier transforms for signals from the medial condyle during extension.

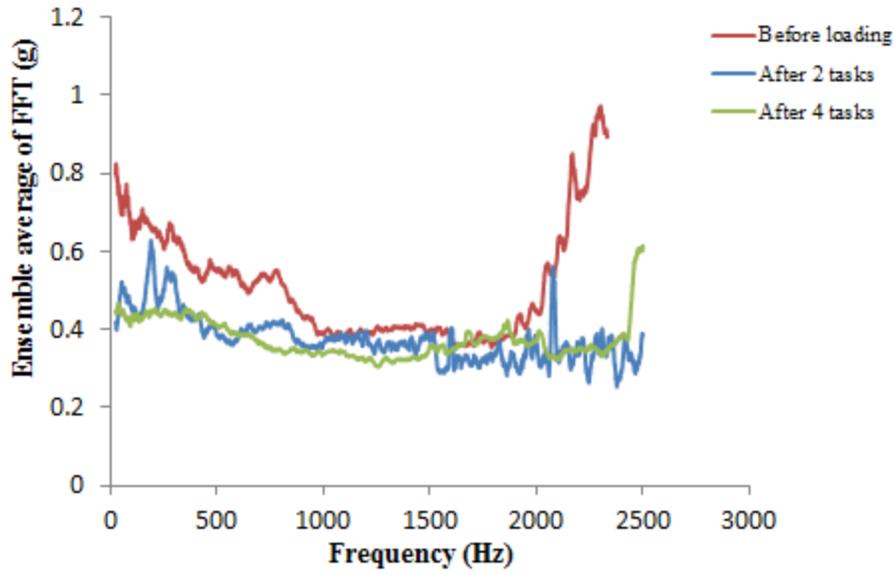


Figure 4.1(b): Ensemble average of the Fourier transforms for signals from the medial condyle during flexion.

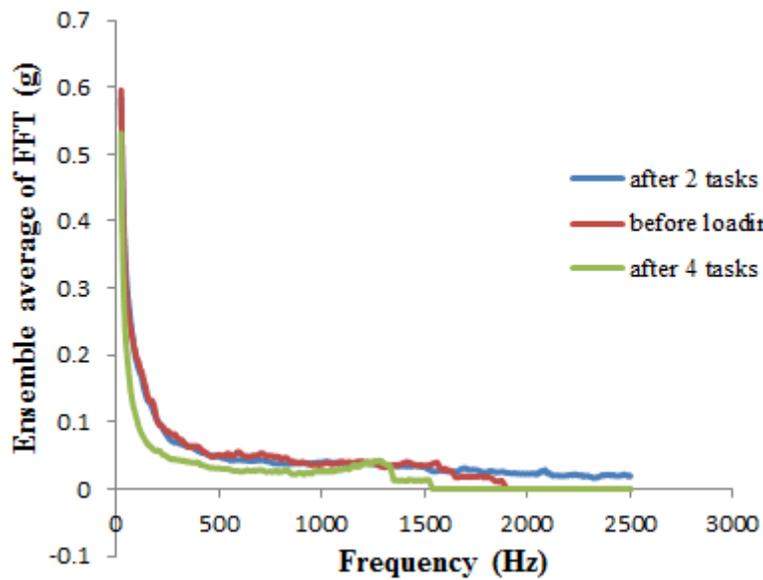


Figure 4.1(c): Ensemble average of the Fourier transforms for signals from the patella during extension.

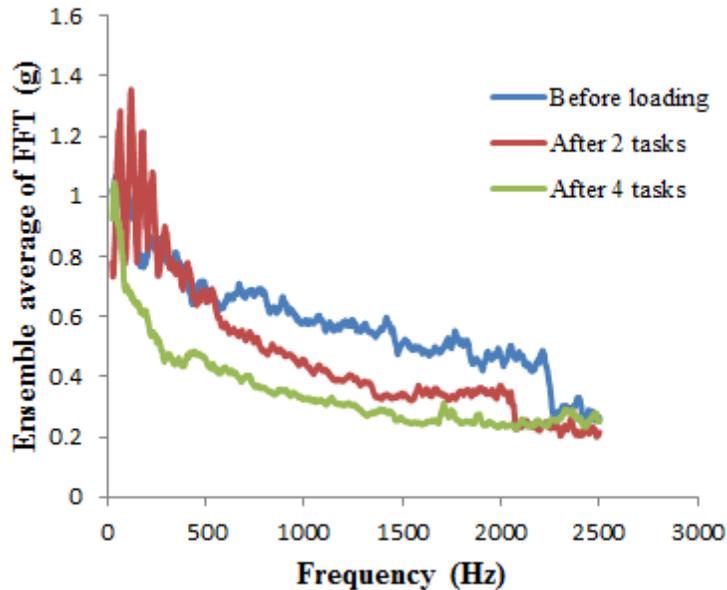


Figure 4.1(d): Ensemble average of the Fourier transforms for signals from the patella during flexion.

Results from t-tests

As it is clear from Figure 4.1 (d), there exists a visually distinguishable difference in the Fourier transforms of the signals in the 750Hz-2000Hz range, across the three different loading conditions. The RMS and median frequency of the signals were computed in this range. A paired, two-tailed t-test did not reveal any significant results in the median frequency whereas the mean RMS power of the signals showed a 40% decrement from pre-test, before loading ($M = 0.70\text{g}$, $S.D = 0.40\text{g}$) to post-test, after 4 tasks ($M = 0.42\text{g}$, $S.D = 0.16\text{g}$) at $t(9) = 2.57$, $p = 0.03$. Specifically, there was a cumulative or time-dependent effect of joint loading on the RMS power of knee vibrations. Table 4.1 (a) and (b) list the values computed for RMS and median frequency of the vibration signal between 750Hz and 2000Hz from the patella, during flexion.

Table 4.1(a): RMS values for vibration signals in the range 750Hz-2000Hz, from the patella during flexion.

	Before Loading			After 2 tasks			After 4 tasks		
	Trial 1	Trial 2	Trial 3	Trial 1	Trial 2	Trial 3	Trial 1	Trial 2	Trial 3
RMS	1.59g	1.60g	1.50g	0.79g	0.79g	0.64g	0.71g	0.49g	0.45g
Mean (SD)	1.57g (0.041)			0.745 g (0.086)			0.549g (0.139)		

Table 4.1(b): Median Frequency values for vibration signals in the range 750Hz-2000Hz, from the patella during flexion.

	Before Loading			After 2 tasks			After 4 tasks		
	Trial 1	Trial 2	Trial3	Trial 1	Trial 2	Trial3	Trial 1	Trial 2	Trial3
Median Frequency	1349Hz	1380Hz	1379Hz	1187Hz	1138Hz	1156Hz	1348Hz	1136Hz	1242Hz
Mean (SD)	1369Hz (17)			1161Hz (24)			1242Hz (106)		

Analysis of Variance

An analysis of variance (ANOVA) performed across the three loading conditions to investigate effects of cumulative load on the RMS of the signal did not reveal any significant results. Therefore, the relative RMS, i.e. RMS of the signal normalized to its initial value (as defined in chapter 3) was considered. A one way within subjects ANOVA conducted to compare the temporal effect of load on the relative RMS of signals before the tasks, after 2 tasks and after 4 tasks resulted in an overall 28% decline in the relative RMS after 4 tasks at $F(2,27) = 3.14$ at $p = 0.06$. From this result we observe that there are no main effects of loading duration on the power of joint vibrations. However, a p -value of 0.06 is very close to the *a priori* α -value of 0.05. This implies that although there are no main effects of load, the data certainly shows a declining trend in the power of vibrations with increase in cumulative load. The significant pre to post decline in absolute power as noted earlier by a t-test, adds value to the observation of such a trend in the power of vibration signals. Figure 4.2 is a graphical representation of the decline in relative RMS power of vibrations with increasing number of tasks performed.

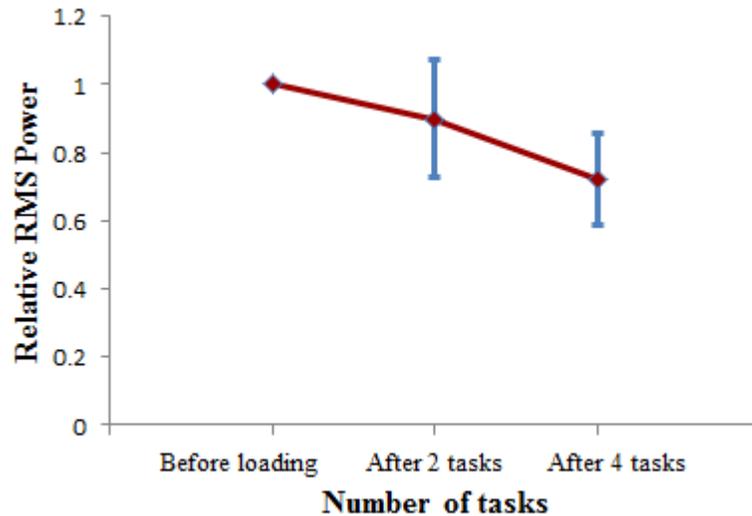


Figure 4.2: Graph displaying the mean and standard deviation of the relative RMS power across all participants, at different points in time based on the number of tasks accomplished. The decrease in the mean relative RMS power is apparent after the participants accomplish two tasks and all four tasks.

4.2 Regression analyses

The results of linear regressions performed between the 3-D cumulative moment and the corresponding relative median frequency (RMF) defined previously, are presented in Table 4.2 below. The results revealed significant correlations (assuming a 95% confidence level), between the induced cumulative load and corresponding RMF for the following cases:

- 1000-2500Hz Patella Flexion: A t-test performed for RMF before loading and after 4 tasks showed a significant difference with $p < 0.0001^*$.
- 20-100Hz Patella Flexion: A t-test performed for RMF before loading and after 4 tasks showed a significant difference with $p = 0.02^*$.

Regression analyses performed between the same variables in the 500Hz-100Hz range did not find any consequential results. Likewise, the RMS analysis did not show any correlations in all the bandwidths considered.

Table 4.2: Results of all regressions performed for Cumulative Load (Nm.s) versus RMF.
 * indicates statistical significance.

Frequency Range		t-test (β , <i>dof</i> , <i>p</i> -value)	R ²	RMSE
		$t(0.95, 49) = 2.01$		
20-100Hz	Patella Extension	$\beta = -2649, p = 0.99$	9.0e-7	11799
	Patella Flexion	$\beta = 618321, p = 0.02^*$	0.1112	11123
	Medial Condyle Extension	$\beta = -78039, p = 0.78$	0.0001	11790
	Medial Condyle Flexion	$\beta = -220858, p = 0.26$	0.0258	11646
100-500Hz	Patella Extension	$\beta = 47431, p = 0.36$	0.0172	11697
	Patella Flexion	$\beta = 8897, p = 0.83$	0.0009	11793
	Medial Condyle Extension	$\beta = -68403, p = 0.57$	0.0067	11759
	Medial Condyle Flexion	$\beta = -1073, p = 0.98$	1.5e-5	11799
1000-2500Hz	Patella Extension	$\beta = -56773, p = 0.43$	0.0131	11721
	Patella Flexion	$\beta = -253667, p < 0.0001^*$	0.2992	9877
	Medial Condyle Extension	$\beta = 8833, p = 0.90$	0.0003	11797
	Medial Condyle Flexion	$\beta = 7310, p = 0.88$	0.0004	11796

Sample means and standard deviations gathered for RMS and median frequency in the 1000-2500Hz range for signals from the patella during knee flexion of one participant are presented in Tables 4.3 (a) and (b).

Table 4.3 (a): Mean median frequency (with Std. Dev) computed for signals recorded after each task from the patella, while under knee flexion.

Task	0	1	2	3	4	5	6
Average Median Frequency (S.D)	1509.82 Hz (6.36)	1492.0 Hz (20.73)	1489.75 Hz (25.62)	1504.81 Hz (6.69)	1521.39 Hz (37.21)	1475.11 Hz (50.96)	1489.65 Hz (5.54)

Table 4.3 (b): Mean RMS power (with Std. Dev) computed for signals recorded after each task from the medial condyle, while under knee flexion.

Task	0	1	2	3	4	5	6
Average	20.45g	20.41g	20.34g	20.29g	20.31g	20.34g	20.32g
RMS (S.D)	(0.04)	(0.007)	(0.02)	(0.03)	(0.02)	(0.16)	(0.02)

The RMF of the signals recorded in the 1000Hz-2500Hz range from the patella during knee flexion declined linearly with increase in cumulative load at $p < 0.0001$. The coefficient of determination (R^2) was equal to 0.29 implying that 30% of the variability in the vibration parameter RMF was explained by the 3-D cumulative joint moment by a linear relationship. The regression plot for this case with individual scatters for each participant is shown in Figure 4.3(a). The residuals for this regression returned a scatter plot which was not entirely random or normal as seen from Figure 4.3(b).

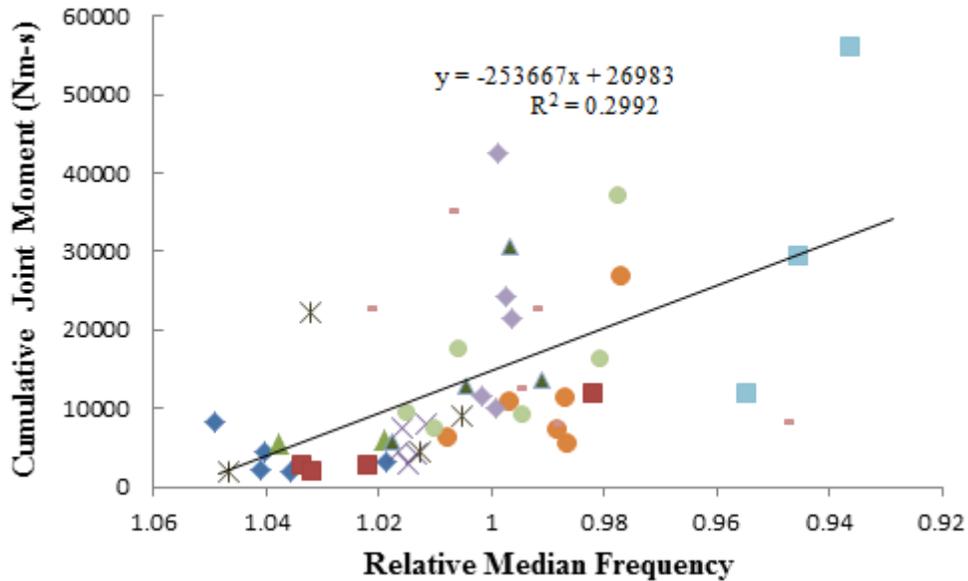


Figure 4.3 (a) : Regression plot of Cumulative moments versus Relative Median Frequency for vibration signals from the Patella in the range 1000-1500Hz during flexion showing individual scatter of points for each participant. Each unique combination of color and shape represents a participant.

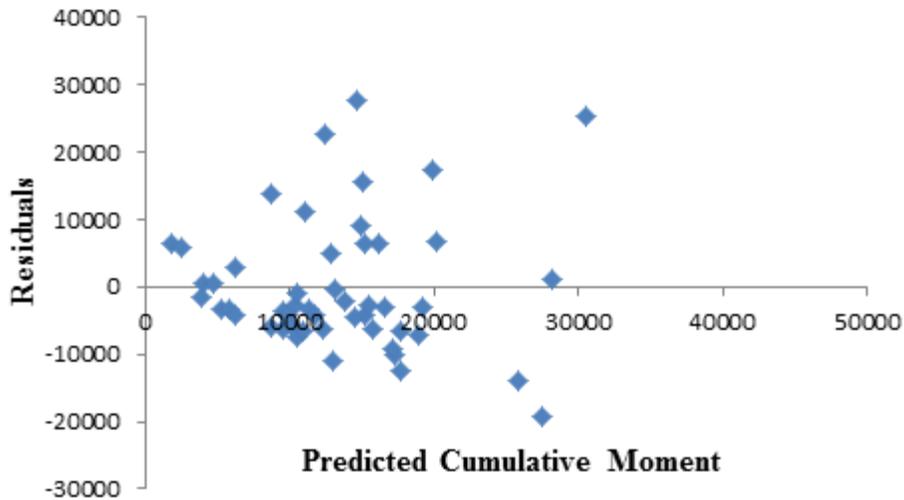


Figure 4.3 (b): Residual plot of linear regression between 3-D cumulative moment and RMF, showing a random, but non-normal scatter of points.

As the residual plot revealed a non-normal distribution, the logarithmic value of the 3-D cumulative moment was computed. A regression plot of this log moment versus the RMF for the 1000Hz-2500Hz patella signal during knee flexion showed an increase in the coefficient of determination by about 3% as seen in Figure 4.3 (c). This means that the log moment explained variability in RMF better than the cumulative moment *per se* by about 3%.

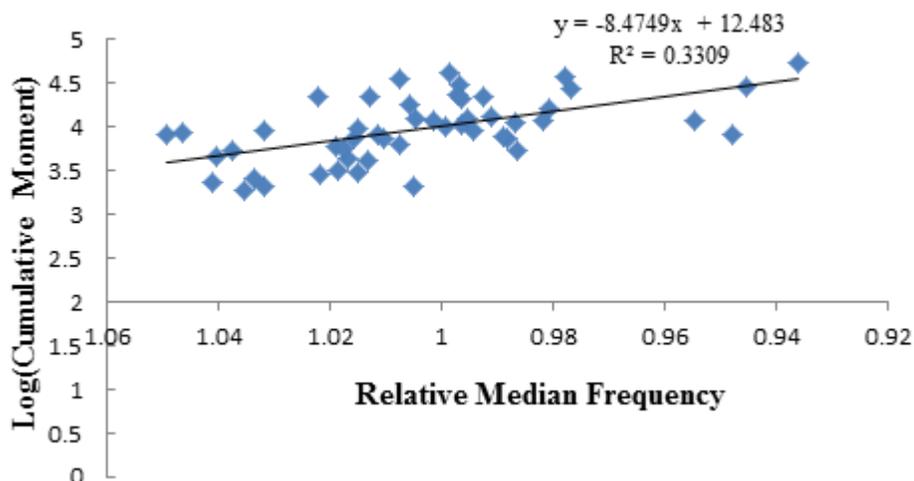


Figure 4.3 (c): Regression plot of the Log of 3-D Cumulative Moment at the knee versus the Relative Median Frequency RMF for 1000-2500Hz range. The coefficient of determination R^2 is

now 33% as opposed to 30% when the regression was performed without the logarithm being taken.

As seen from Table 4.2, linear regression analysis performed between the cumulative moments and RMF of signals in the 20Hz-100Hz range captured from the patella during knee flexion showed a declining association, at $p = 0.02$ and with $R^2 = 0.11$. This means that the 3-D cumulative moment decreased linearly with decline in the vibration parameter RMF, explaining about 11% of the variability in RMF. The regression plot for this case with individual scatters for each participant is shown in Figure 4.4(a). The residuals for this regression returned a non-normal scatter plot as seen from Figure 4.4(b).

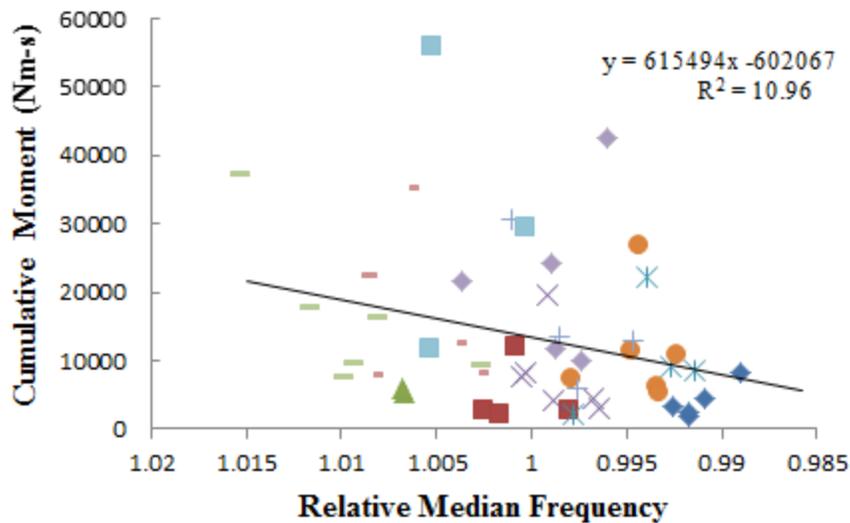


Figure 4.4 (a) Regression plot of Cumulative moments versus Relative Median Frequency for vibration signals from the Patella in the range 20Hz-100Hz during flexion showing individual scatter of points for each participant. Each unique combination of color and shape represents a participant.

As the residual plot revealed a non-normal distribution, the logarithmic value of the 3-D cumulative moment was computed. A regression plot of this log moment versus the RMF for the 20Hz-100Hz patella signal during knee flexion showed an increase in the coefficient of determination by about 2% as seen in Figure 4.3 (c). This means that the log moment explained variability in RMF better than the cumulative moment *per se* by about 2%.

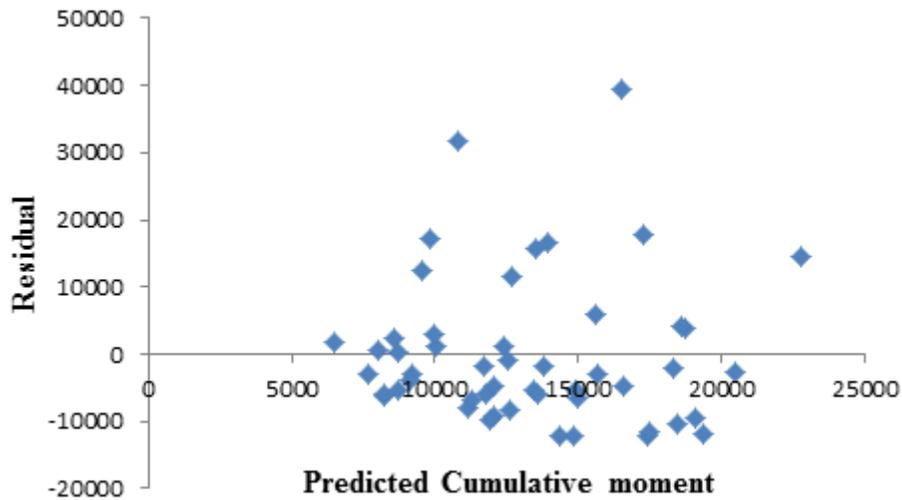


Figure 4.4 (b): The residual scatter plot for the linear regression performed between 3-D cumulative load and the RMF for vibration signals in the 20Hz-100Hz range recorded from the patella during knee flexion.

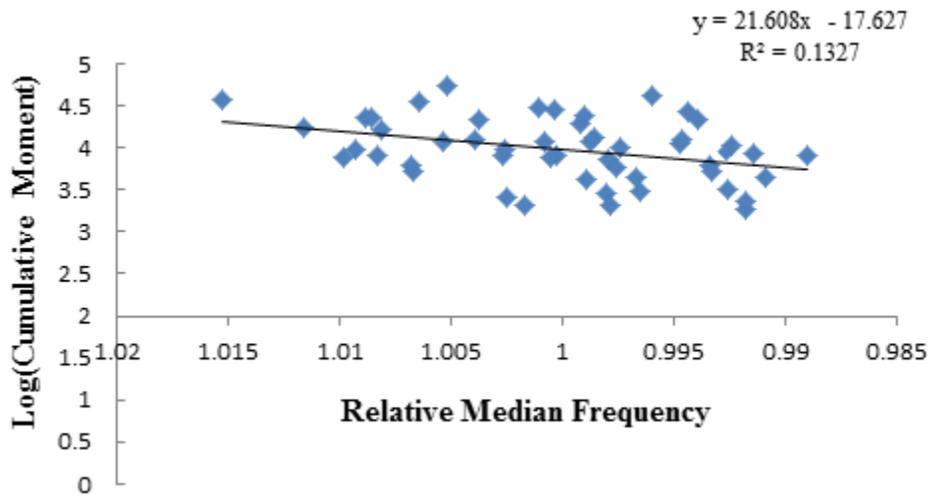


Figure 4.4 (c): The regression plot of the log of 3-D cumulative moment versus vibration signal RMF in the range 20Hz-100Hz captured from the patella during knee flexion.

4.3 Cumulative moment normalized with respect to body mass.

The cumulative load on the joint is influenced by the participant's body mass among other variables (Norman et al., 1998). Therefore the cumulative load was normalized with respect to body mass and this normalized cumulative moment (Nm.s/kg) was regressed with the spectral parameter, RMF for the 1000-2500Hz Patella Flexion and 20-100Hz Patella Extension cases. The linear regression performed between the normalized cumulative moment and the Relative Median Frequency for signals between 1000Hz and 2500Hz resulted in an increase in the coefficient of determination (R^2) by 1%. The R^2 value for the new regression was 0.30. There was no increase in R^2 for the 20-100Hz Patella flexion case.

4.4 Two-factor regression model

Since two significant but opposite associations were observed, a two factor regression model was considered, with the predictor variables as the RMF in the two conditions in which opposite associations were observed. The dependent variable was set as the cumulative moment normalized to body mass and the resulting model returned an R^2 value of 32.27%, which was higher than the individual R^2 values and the coefficients and p-values for each variable are presented in Table 4.3. Figure 4.5 (a) shows the scatterplot of the actual versus predicted cumulative moment.

Table 4.3: p-values for the two predictor variables of the two-factor model for signals captured from the patella during knee flexion

Term	Estimate	Std. Error	t Ratio	Prob > t
20-100Hz (X_2)	3979.76	3338.57	1.19	0.2392
1000-2500Hz (X_1)	-3232.99	835.11	-3.87	0.0003*
Intercept	-550.73	3713.63	-0.15	0.8827

The two-factor model establishing the estimated value of the normalized cumulative moment as a function of the two predictor variables which are RMF of the signals in the 20Hz-100Hz range and the 1000Hz-2500Hz range is:

$$\bar{Y} = -550.73 - 3232.99X_1 + 3979.76X_2$$

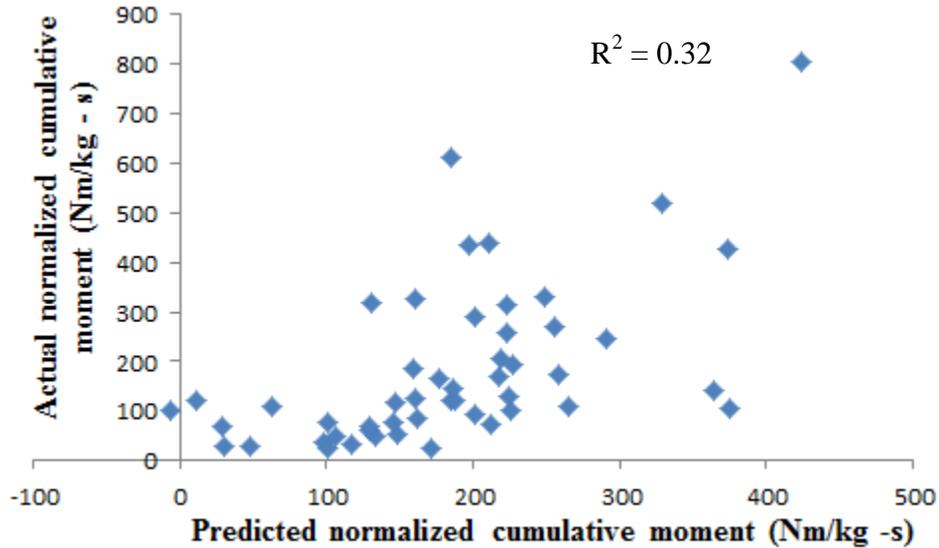


Figure 4.5: Scatterplot of actual versus predicted values of the cumulative moment normalized to body mass.

As seen from the scatter plot, $R^2 = 0.32$. Thus, the two-factor model shows a greater correlation than each of the individual factors. Also, from Table 4.3 we see that the signals in the 1000Hz-2500Hz range exhibit a significant influence on the model at $p = 0.0003$. Hence, these signals are more robust compared to the 20Hz-100Hz signals.

Multicollinearity

The two predictor variables were also regressed upon each other to test for multicollinearity and the association turned out significant ($p = 0.0136$). This means that the two factor model is not necessarily very robust because of the presence of an association between the two predictor variables. However, the degree of association, R^2 for this plot turned out to be low at about 12%

and typically R^2 values less than around 16% do not affect the variance in the model due to multicollinearity significantly. Gravel et al. (1997) who developed predictive equations for maximal static moments generated by trunk musculatures disregarded one of their regression cases due to low R^2 values of 0.09-0.12. Moreover, the two-factor model is exploratory and has been created from a small sample size of about ten participants. A larger sample size should be investigated before anything can be confirmed regarding the multi-factor model.

5.0 DISCUSSION

The purpose of this exploratory study was to investigate the effect of mechanical loading on the vibration properties of the knee joint. Preliminary analyses of the knee joint signals showed that the vibration properties including RMS power and relative median frequency of the signal significantly change with increase in cumulative load on the knee. Specifically, signals from captured from the patella during knee flexion showed these differences. The results of the study indicate the possibility of a relationship between joint exposure to load and vibration signals emitted during joint movement. However the quantitative nature of this relationship is not clear yet although the current study points towards a nonlinear model.

Two types of analysis were conducted to study vibrations. First, RMS and median frequency of the vibration signals captured from the patella during knee flexion, by a piezoelectric accelerometer were computed in the 750Hz-2000Hz range. This bandwidth was based on results obtained from the ensemble average of the participants' frequency spectra for various vibration signals and loading conditions. Second, the signal RMS and relative median frequency were computed in the ranges 20Hz-100Hz, 100Hz-500Hz, 1000Hz-2500Hz using the Butterworth filter. These bandwidths were determined based on pilot studies and literature review of pathological vibrations.

5.1 General findings

In the first analysis, a pre-post t-test revealed that the RMS of the signal from the patella under flexion, declined with increase in loading. This leads us to reject the null hypothesis 1a, that there is no difference in the RMS of the signal before and after loading the joint. However, contrary to our expectation as outlined in alternate hypothesis 1a, the RMS decreased as the cumulative load increased and this leads us to question the etiology of vibration signals from the knee. A possible explanation for this phenomenon of decline in RMS could come from understanding change in cartilage morphological characteristics, such as cartilage thickness and area in relation to vibration signals from the joint. From an ANOVA, we observe a declining trend in the normalized power of vibrations with increasing cumulative load. As such, we fail to reject the null hypothesis that there exists no relationship between the vibration characteristic RMS power and cumulative load. However, since a trend was observed in the power of

vibrations, the possibility of a relationship between vibration characteristics and cumulative load should be explored further.

In the second analysis, the relative median frequency of vibrations captured from the patella during flexion in the frequency range of 1000Hz-2500Hz declined while increasing the cumulative exposure on the joint. The coefficient of determination R^2 for the linear regression between the relative median frequency and the 3-D cumulative moment turned out to be nearly 30%. This implies that about a third of the variability in cumulative exposure could be explained by the vibration characteristic, relative median frequency. These findings suggest a quantitative relationship between the variables, leading us to reject the null hypotheses 1b and 2b. Moreover, this also means that we fail to reject the alternate hypothesis 1b and 2b that the median frequency decreases with increase in exposure since we see a declining frequency characteristic as purported. As such, we fail to reject the null hypothesis that there exists no relationship between the vibration characteristic RMS power and cumulative load. However, since a trend was observed in the power of vibrations, the possibility of a relationship between vibration characteristics and cumulative load should be explored further.

5.2 Frequency range under consideration

One of the purposes of constructing the ensemble average of participants' frequency spectra was to gather an understanding of the frequency bandwidth in which the vibrations across loading conditions varied from each other. By visual inspection of the ensemble graphs, the 750Hz-2000Hz bandwidth for vibrations under flexion from the patella were most distinguishable in terms of magnitude across loading conditions. Since the study was performed on healthy adults, this finding was indirectly supported by McCoy et al. (1987) who asserted that pathological vibrations in the knee exist only at frequencies below 1000Hz. They used miniature accelerometers affixed on the patella and condyle and suggested that pathological vibrations due to meniscal lesions exist below 100Hz and those due to signs of degeneration between 300-600Hz. Rangayyan et al. (1988) collected joint vibration signals from the patellofemoral joint using miniature sensors and determined that mild chondromalacia existed at higher frequencies (up to 300 Hz) than severe chondromalacia which showed vibrations below 100Hz.

These studies added evidence to the notion that pathological knee vibrations existed only below 1000Hz and basically led to the proposition in hypothesis 1b that the frequency of vibrations decreased as severity of joint deformation increased. Since our study was conducted only on individuals with no history of joint pathology, it is likely that vibrations from these healthy joints would be composed of higher frequencies than vibrations from pathological joints. Besides, in our second analysis, the relative median frequency found changes in the 1000Hz-2500Hz bandwidth. This is again in tandem with the proposition that the frequency of the joint vibrations would be greater for healthier joints as compared to pathological joints.

5.3 Individual differences in vibration behavior due to diurnal changes and time-dependent joint deformation.

The scatter of the relationship between the relative median frequency and cumulative 3-D joint moment as seen in Figure 4.2 (a) and Figure 4.3 (a) show a unique trend for each individual. This implies that each person's joint surfaces reacted differently to the squat-lifting tasks assigned. The possible reason for such a distributed scatter across participants could be because joint deformation is based on the existent morphology of the joint and that the effect of daily activity on joints has not been considered. Koo et al (2011) conducted a cartilage MRI study on healthy volunteers and suggested that even walking is an important consideration while analyzing cartilage surface characteristics. As a result, each individual's joint morphology may have been different when they walked into the lab for the experiment and due to the viscoelastic nature of joint tissues, each joint may have responded differently to mechanical stress. This means that there could have been carry-over effects of loading on the joint, thus causing individual variations in the vibration responses. Although in the current study we tried to partially account for these deformational changes by normalizing the signal parameter relative to its initial value, carry over effects in case of the complex cartilage behavior cannot be discounted completely. This phenomenon of carry-over effects due to diurnal changes and time-dependent deformation is further explained below.

In the current study, although all the participants were similar in terms of age and fitness routines, the experiment was not performed at the same time of the day for everyone. As a result, the diurnal variation in physical activity levels was not monitored in the study. Eckstein et al. (1998) conducted experiments where participants with healthy knees performed 50 knee bends

for a few minutes and MRI images of the cartilage were reconstructed before and after the exercise. They discovered a 6% mean decrease in cartilage volume after 3-7 minutes of physical exercise and a 5.2% mean decrease after 8-12 minutes of exercise. Their suggestion was to allow patients an hour of rest before recording the MRI so as to avoid effects of daily activity.

Likewise, there can be similar deformational changes in the knee over the course of a day due to routine physical exercise. Due to constraint on lab schedules, the experiment was performed at different times of the day and participants' activity levels for hours prior to the experiment could not be controlled because of participants' work commitments. As a result, each participant probably had a different joint condition when they came in.

Hosseini et al. (2010) discovered that the cartilage deformation was time dependent and that for healthy knees, cartilage deformed rapidly for the first 20s and the rate became constant after about 50s. This implies that deformation behavior is dependent on the existing condition of the cartilage. Such carry over effects can also be due to a variety of factors as joint response behaviors could differ across ethnicities, genders and age. However, in this experiment, only age and exercise patterns have been controlled for. Gender and ethnicity effects have not been studied. Thus individual differences due to daily activity could have been a major cause for the individual scatters of figures 4.2 (a) and 4.3 (a).

5.4 Only the signals captured from the patella during passive movement of limb were consequential.

An interesting observation in this study was that only signals captured from the patella showed useful results. The reason could be that there is a lot of intermediary skin at the medial condyle gap for vibrations to be recorded without attenuation by skin friction noise. At the mid-patella, the accelerometer comes directly in contact with the patella cap and as per Rangayyan et al. (2000), all the vibrations inside the joint travel to the patella cap.

Another interesting observation was that all significant findings were only in cases of knee flexion when the limb moved passively as opposed to extension when the quadriceps and shank muscles are not actively engaged. This is in tandem with literature (Reddy et al., 1995) which stated the use of passive limb movement for producing significant results while recording joint vibrations. This finding is also supported by Rangayyan et al. (2000) who proposed that muscle contractions can contribute to noise in vibration signals. However, they used advanced signal

processing to eliminate the contractions as opposed to passive limb movement which is a simpler choice.

5.5 Possibility of a non-linear or multi-variate relationship.

There are several points in the results of this study which indicate the possible non-linearity of the variables under consideration. Firstly, a log transformation of the 3-D moment gave a stronger association as opposed to using just the 3-D moment in the linear regressions. Secondly, the residuals for the significant linear regressions were non-normal indicating that a linear plot may not be the ideal regression curve. Thirdly, despite a small sample size the ANOVA result between the relative RMS power and time indicates a declining trend in the RMS parameter with increase in cumulative load over time, thus suggesting the possibility of a relationship. Further studies should account for other risk factors which could strengthen this trend in signal power in order to find out the nature of the potential relationship.

The results of linear regression between cumulative moment and RMF in 20-100Hz bandwidth showed an R^2 value of 11% while the 1000-2500Hz bandwidth had an R^2 of 29%. Therefore with these conditions as predictor variables, a combined two factor model was investigated and this returned a higher coefficient of determination (R^2) of 32% than the individual R^2 values. Thus, the two factor model seemed to have better predictive ability and that in future, the effect of multiple vibration parameters must be examined simultaneously.

6.0 LIMITATIONS AND FUTURE DIRECTIONS

This study was very exploratory in nature and therefore various aspects of the methods applied to answer the research question can be modified to yield stronger associations in the results or to reveal new findings. As the current study was limited by sample size and by various other factors of the variables chosen, much of the future works needs to focus on understanding the fundamentals of the vibration emission phenomenon rather than taking this study to the next level. Some limitations and scope for future work are discussed as follows.

6.1 Etiology of vibrations

Effects of joint velocity

Vibration frequencies below 20Hz have been eliminated to remove skin friction artifacts or snapping of tendons as purported by Chu et al. (1978). Similarly, high frequencies (above 2500Hz) were not even considered as all mechanical knee vibrations discovered so far have been well below that. However there is no definitive proof that the vibrations captured indeed originate from the soft tissues and are not irrelevant noises during joint movement. Joint vibration characteristics can be influenced by kinematic parameters like joint velocity. Kernohan et al. (2006) examined 24 pathological joints at different velocities and found out that the root mean square power of joint vibrations gets significantly affected by changing the velocity of joint movement. Besides, there are some normal joint sounds which are associated with a particular joint angle or joint velocity. For instance, the patellofemoral click (PFP) is a routine phenomenon which is associated with extremely slow movement in healthy joints irrespective of their loading (McCoy et al., 1987). Although the speed of the reference movement used in the current study has been kept constant to 6 seconds/cycle in tandem with studies by Tavathia et al. (1992) and Reddy et al. (1995), these speeds are largely determined by the participants' comfort level and do not necessarily represent an optimum speed of recording the pathological aspect of vibrations. The independent variable (3-D joint moment) considered in the present study pertained to joint kinetics rather than kinematics. Future studies should investigate the effect of kinematic variables such as joint angles and joint velocities on the vibration parameters to check for the possibility of stronger associations.

Changes in joint mechanical properties

Cartilage tissues exhibit senescence, which is a phenomenon where cells of biological tissues stop regenerating. This increases the sensitivity of the cartilage response to joint loading.

Senescence can be induced due to aging and degenerative changes in the cartilage. As a result, a healthy cartilage would be less sensitive to joint loading as opposed to an already degenerating or aging cartilage. The current study used healthy participants making it difficult to induce a deformational change in cartilage response behavior. This could explain why mechanical load measured as the joint moment, explained only about 30% of the variability in vibrations. Many industry workers undergoing squat-lifting on a daily basis are much older than the student population represented in the current study. These workers' joints may actually be more sensitive to changes than those of the students who took part here, by virtue of their age. Future studies should examine participants who are similar to actual factory workers in age and physical conditions, so that the results represent real conditions.

The knee joint is a complex structure composed of different types of tissues and synovial fluid. The philosophy behind the current study has majorly been that vibrations are caused due to friction between the meniscal surfaces and release of viscoelastic strain energy. Since mechanical stress can be a reason for changes in joint mechanical properties, it might be worthwhile to study the viscoelastic creep and relaxation behavior of joints with respect to the joint vibration emission. Understanding this relationship first could provide a more holistic insight into why vibration parameters increase or decrease with joint load.

6.2 Direct measures of loading.

Although the external moment has been used several times in the past to quantify joint loading, it is only driven by external forces such as box weights and body weight and does not include internal compression and shear forces acting upon the joint. The knee joint being a complex structure of soft tissues and fluids, the 3-D cumulative moment alone is not a complete representation of the stresses acting upon the knee. Winter (1995) asserted in his book that co-contraction forces should be imposed on the free body diagram while constructing a biomechanical model. As squat lifting tends towards an extreme posture for the knees, a more holistic approach involving direct compression and shear forces, ligamentous forces, co-

contractions and fluid pressure on the meniscus should be considered while investigating joint loading.

Future studies should therefore consider measuring joint force by methods which directly influence injury such as biomechanical knee models to predict direct compression or shear forces on the meniscus. Electromyography to capture muscle forces and co-contractions should be explored for different muscle groups which influence squatting, including the quadriceps. These studies could potentially reveal a dominant force affiliated with joint deformation.

Mechanomyography from muscle groups of the lower extremity should also be explored for comparison with joint vibrations. By examining measurements which are more directly associated with injury causation, the correlation between joint vibrations and mechanical load may be strengthened, possibly setting the scope for a quantitative relationship.

6.3 Static versus dynamic loading

The squat-lifting tasks performed were dynamic loading tasks. There is little literature which conducted *in vivo* studies, suggesting the dynamic behavior of the cartilage under load. Kääl et al. (1998) examined the behavior of cartilage recovery *in vitro* by subjecting rabbit knee cadavers to static as well as dynamic loading and subsequently examining the morphology of the cryogenically preserved cartilage tissue under an electron microscope. The cadavers were subjected to a static load at 90° flexion for 30 minutes and dynamic load at about 1 Hz frequency from 70° to 150° for 30 minutes. These dynamic parameters are similar to what has been used in our *in vivo* study where the subject was loaded under a frequency of up to 40lifts/min (0.66 Hz) while squat-lifting for nearly 30 minutes. Their results revealed that cartilage deformation in terms of thickness and time of recovery was significantly lesser and faster for dynamic loading than static loading. This directs to the implication that a static load may have induced a higher deformation than dynamic squat-lifting. As some of the industrial squatting tasks (such as in farming) can last for hours through the day, it may be worthwhile to investigate static tasks in the future.

In order to test for the impact of static loads, a pilot was performed where the participant was made to half-squat for as long as he physically could (~2 minutes). No differences in the power spectrum were observed in this pilot. Future studies should however focus on static loading for extended time periods in deep knee flexion.

6.4 Advanced signal processing methods

According to Krishnan, Rangayyan et al. (2000) the joint vibration signal is non-stationary in nature since the quality of joint surfaces in contact changes with the joint kinematics. Therefore, these authors have attempted several advanced methods of signal processing for over a decade in order to extract useful features and parameters from the signal. Due to the elementary and exploratory nature of the current study, signal processing methods applied were restricted to examining the frequency spectrum or time series as a whole rather than applying segmentation techniques. Use of advanced digital signal processing in future, might yield dependent variables which correlate stronger with mechanical loading.

7.0 CONCLUSION

The results of this exploratory study revealed that cumulative mechanical exposure on the joint has an effect on the characteristics of vibration signals emitted by the articulating joint. The RMS power of the signals showed a declining trend in the 750Hz-2000Hz range with increase in mechanical exposure. With the increase in cumulative moment on the joint, the normalized median frequency of vibration signals was found to decrease for vibration signals in the 1000Hz-2500Hz range and increased for signals in the 20Hz-100Hz range. These preliminary findings imply the probability of a statistical relationship between vibration characteristics and joint mechanical exposure. The quantitative nature of this relationship is not known yet although the findings point towards a non-linear, or a multi-variate model. Cumulative moment being an indirect measure of joint stress limits our understanding of this relationship. Therefore as a first step, future studies should address the dependence of joint vibrations on cartilage morphological changes. Also, future studies conducted should be controlled in terms of accounting for exercise, and obtaining a direct means of measuring cumulative load than the 3-D moment.

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APPENDIX A: DEMOGRAPHIC QUESTIONNAIRE

Participant no. _____

Please take some time to answer the following questions. All answers will be kept confidential.

1. Age _____
2. Gender : Male _____ Female _____
3. Height : _____
4. Weight : _____
5. Occupation : _____
6. Have you had any ankle, knee or hip injury in the past? Yes _____ No _____
If yes, please explain _____
7. Have you experienced injury on any other part of your body for which you had to take days off from work/school? Yes _____ No _____
If yes, please explain _____
8. Do you experience any discomfort while squatting on the floor?
Yes _____ No _____ Don't Know _____
If yes, please explain _____
9. Do you have any condition which restricts you from physical activity?
10. Dominant Leg _____.
11. How many times do you exercise per week? _____. What kind of exercise?_____.

-----Thank you for your time!-----

APPENDIX B : CONSENT FORM

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants in Research Projects Involving Human Subjects

Title of Project

Quantification of cumulative load on the knee using a vibration emission method.

Investigator(s)

Principal Investigator: Dr. Michael J. Agnew

Co-Investigator: Navaneeta Dorbala

I. Purpose of this Research/Project

The purpose of this study is to measure loads on the knee induced by various occupational tasks which involve squatting. This shall be done by capturing acoustic signals emitted from the movement of the knee joint.

When two surfaces with some wear and tear slide against each other, acoustic energy in the form of vibrations is dissipated due to the friction between them. Similarly, the surfaces of the knee joint which facilitate body movement undergo wear and tear during daily activities and dissipate acoustic energy. These surfaces are manifested in the form of the articular cartilage, or just cartilage. Sliding movements between the rough surfaces of the cartilage cause pain in the knee. In many sports such as running, soccer, football, basketball and tennis, players use the knee as their primary tool for twisting, rotating and for bearing several times their body weight. As a result, these sportsmen are prone to excessive loading on their knees, resulting in a rough cartilage which pains during certain movements. The acoustic signature of joint tissues, particularly the cartilage, has been used in the past to distinguish between a healthy and a diseased knee.

Since the knee joint is the primary weight bearing joint, the nature of a task can affect the quality of the surface of its cartilage. Several occupations including construction, mining and agriculture involve performing tasks which require frequent squatting while simultaneously lifting and

lowering heavy weights with no adequate rest.

Squatting is a restricted and awkward posture which can cause a lot of mechanical stress on the knee joint. Squatting with lifting increases the load placed on the knee by several times. Joint tissues such as the cartilage have limited tolerance for withstanding mechanical loads. So just like any other material, their threshold of failure decreases if they are under forceful and repetitive loads without rest. Such loads are called cumulative loads and they can place heavy stresses on the knee joint causing wear and tear in the cartilage tissues. Therefore, injuries and illnesses of the knee are commonly reported among workers in such occupations. Our goal is to quantify wear and tear in cartilage tissues and associate this with the cumulative load exposure caused by the industrial task.

Cumulative loading on the knees should be minimized in order to avoid the occurrence of injury or illness due to occupational tasks. As a first step, it is important to quantify this load. In this study, cumulative load on the knees while squatting shall be quantified in terms of the acoustic signature of the cartilage. A statistical model elucidating the impact of various loads on changes in the acoustic signature This will hopefully broaden our understanding of the impact of industrial tasks in restricted postures on the health of the joint tissues. Furthermore, it may be possible to use the “acoustic emission” method as a portable and easy means to measure occupational loads in field settings rather than lab settings. Moreover, quantifying cumulative load on the knee can be useful for estimating the long term effect of knee loading among workers.

Twenty people between the ages of 18 and 30, reflective of the university community shall participate in this study. All participants chosen will be physically and mentally strong to perform in the study without any history of joint disorders.

II. Procedures

As a participant in this experiment, you will be expected to attend one session of testing, lasting from 1-2 hours in the Industrial Ergonomics and Biomechanics Laboratory at 5th floor, Whittemore Hall, Virginia Tech. During the session, you will be asked to perform the following

procedures.

Throughout the experiment, you will have tiny markers taped to various joints on your body for recording your movement data.

There will be several lifting tasks to perform. You will be asked to squat and lift dumbbells in pairs of 1lb, 5lbs and 10lbs. Each of these tasks shall be done repetitively at two different speeds. Thus, there are in total six tasks –i.e., combinations of three weights and two speeds. Each task shall be performed for about 8 minutes, with rest breaks allotted. Thus, the exercise would last effectively for a period of 50 minutes to 1 hour excluding rest breaks.

Measurements will be taken every 8-10 minutes, after completion of each task. For measurements, you will have a tiny accelerometer attached to bony prominences on your knee. This shall be used to collect acoustic signals from the knee. For collecting this data, you will be asked to swing your leg according to a certain cadence. This is our reference movement and acoustic data recorded in this movement after each task , shall be used for analysis.

If you experience any discomfort at any point of time during the experiment, you are completely free to let the researcher know that you want to stop. You may then leave.

III. Risks

You might experience some muscle fatigue while doing these tasks. This will be the same type fatigue you may experience when performing day to day activities, such as climbing stairs, lifting your backpack from the floor or lifting a child. There might be mild muscle soreness for the next 24-48 hours or so. You are free to quit at any point of time in the experiment.

IV. Benefits

The methods and results of this study can be of use for early assessment of cartilage health among workers in mining, agriculture and construction who have to frequently adopt awkward postures due to the nature of their work. This can also be useful to predict exposures due to a certain job over a period of time.

In sports, many players exert several times their body weight on the knees during sporting actions such as bowling, kicking, twisting or throwing. As a result, players are prone to a condition called chondromalacia patellae which is pain due to roughening of the cartilage surface. This technique offers a method of cartilage health assessment among players of sports such as football, basketball and cricket.

Such a technique can be used by clinicians in an easy and affordable manner to assess a person's day-to-day mechanical exposure levels and cartilage health resulting from routine tasks such as climbing the stairs, cleaning the floor, working in the kitchen or carrying children. The earlier the risk of such exposure is known, the easier it is to provide recommendations for better designs at home and work in order to avoid the occurrence of chronic osteoarthritis. Subjects may contact the researcher at a later time for a summary of the results if interested.

V. Anonymity and Confidentiality

All participant files shall be identified by a participant number. The files will be stored as a soft copy which will be accessible only to the student researcher and the principal investigator in charge of this project. No information of the participants and the results shall be shared without the written consent of the participant. The files will be destroyed once the study has been approved for publication in a journal.

VI. Compensation

Participation in this experiment is completely voluntary and no compensation shall be provided for it.

VII. Freedom to Withdraw

At any point of time during the experimental session, participants are free to withdraw at their will if they begin to feel any discomfort due to pain or physical fatigue.

VIII. Subjects' Responsibilities

I voluntarily agree to participate in this study.

IX. Subject's Permission

I have read the Consent Form and conditions of this project. I have had all my questions answered. I hereby acknowledge the above and give my voluntary consent:.

_____ Date _____
Subject signature

_____ Date _____
Witness (Optional except for certain classes of subjects)

Should I have any pertinent questions about this research or its conduct, and research subjects' rights, and whom to contact in the event of a research-related injury to the subject, I may contact:

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