行政院國家科學委員會專題研究計畫成果報告

膝關節受衝擊時之力學與破壞分析 –
退化性與外傷後關節炎形成之物理機制探討

計畫編號：NSC91-2320-B-002-159

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由於本計畫之執行，已有三篇之研討會論文與一篇之期刊論文發表，成果堪稱豐碩。以下是論文目錄以及內容。

研討會論文

1. **Wang JL,** Chung CH, Chiang CK, How the external impact energy affects the internal kinetics of knee joint? The comparison of porcine and human knee joint, ASME International Mechanical Engineering Congress and R&D Expo (IMECE), 2003, Paper No. IMECE2003-42913.

2. Chung CH, **Wang JL**, Chang CH. The effect of loading energy, shock absorber and knee joint postures on the condyle contact pressure during compressive impact loading – an in vitro biomechanical porcine model. Submitted for International Conference on Mechanics in Medicine and Biology (ICMMB), 2003.

3. Chang CH, **Wang JL**, Chung CH. The acceleration attenuation properties of porcine, intact human and human knee joint after meniscectomy during compressive impact loading, Submitted for International Conference on Mechanics in Medicine and Biology (ICMMB), 2003.

期刊論文

1. Wang JL, Lee YL. The shock attenuation properties of straight standing knee joint during different shock absorbers and energy input. Journal of the Chinese Institute of Engineers, 2003, In Press, *Corresponding Author, (SCI)

限制與未來的展望

由於本計畫僅資助一年，因此無法完成第二階段之活體動物實驗，希望將來能使用活體膝關節做測試。

Proceedings of IMECE'03
2003 ASME International Mechanical Engineering Congress & Exposition
Washington, D.C., November 16-21, 2003
IMECE2003-42913

HOW THE EXTERNAL IMPACT ENERGY AFFECTS THE INTERNAL KINETICS OF KNEE JOINT? THE COMPARISON OF PORCINE AND HUMAN KNEE JOINT

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ABSTRACT

Degenerative osteoarthritis is recognized as the consequences of mechanical injuries. The abnormal impact force applied to articular cartilage would result in bone fracture or surface fissuring, and would cause the osteoarthritis [1,2]. The relation among the injury and impact energy was well studied. However, how the external energy attenuated to the internal joint is not carefully studied yet. The porcine knee joint was used as a biomechanical model for the simulation of human knee joint during impact loading. The objective of current study was to find the variation of kinetic characteristics between human and porcine knee joint during axial impact loading. Eight fresh-frozen knee joints from 10 month-old swine and seven cadaver human knee joints were used in the experiment. The mechanical responses such as forces and bending moment of knee joint, and the accelerations of femur was quantitatively analyzed. The results showed that the axial force response between human and porcine joints was similar, however, the anteroposterior shear, flexion bening moment and accelerations of these two joints were different.

MATERIALS AND METHODS

Eight fresh-frozen porcine and seven human knee joints were used. In preparation for the testing, the specimens were stripped of all soft tissue except at the knee joint, where all ligamentous, capsular, and intracapsular structures were preserved. The specimens were than potted with quick setting epoxy at both ends. A "drop-tower type" impact testing apparatus was used for the testing (Figure 1). An impactor guided by two rods was dropped from the top of the tower to produce the impact energy. The energy was transmitted to the specimen through the impounder. The shock absorber was mounted on the top of the impounder. The shock absorber was placed on the top of the impounder to control the impact contact period. The shock absorbers are able to give, approximately, the impact contact time at levels of 20 (stiff), 40 (medium) and 60 (soft) mini-seconds. The specimen was mounted vertically below the impounder and above the six-axial force load cell (AMTI MC6-6-4000, Advanced Mechanical Technology, Inc., Watertown, MA, USA) to find the axial force, anteroposterior shear, and bending moment responses. One two-axial accelerometer (ADXL250EB, Analog Device Inc., USA) was placed at lateral side of distal femur to detect the axial and anteroposterior accelerations. The experimental protocol includes three factors (damper, rotation, and alignment), and each factor has three levels. Every specimen was tested at three different shock absorbers. The posture of knee joints includes neutral, 10˚ inner and outer rotation, and 3.4˚ varus and valgus. The impact weight is 12 kg. The impact height is 30 mm for human joints, and 30, 40, 50 mm for porcine joints; hence the impact energy input is 3.6 J for human joints, and 3.6, 4.8, 6.0 J for porcine joints. The Box-Behnken experimental design was used, hence 14 tests were done for each sample. 230 tests in total were conduct. Signals of reaction force and accelerations were recorded at 10 kHz sampling frequency. The signals were than low pass filtered at 500 Hz frequency using Butterworth filtering algorithm. The following mathematical model was build to find the relationship between the dependent variables with respect to the impact contact time.

$$Y = \beta + (\alpha - \beta)e^{-t/\tau}$$

The least-squared curve fitting method was use to find the constants α, β and τ with the constraint that β must be greater or equal to zero. The correlation of coefficient between experimental data and model was also calculated to find the goodness of fit of the current model. The Matlab optimization toolbox (MATLAB 5.2, Mathworks, Inc., Natick, MA, USA) was used to find the coefficient of the regression equations and the correlation coefficient.

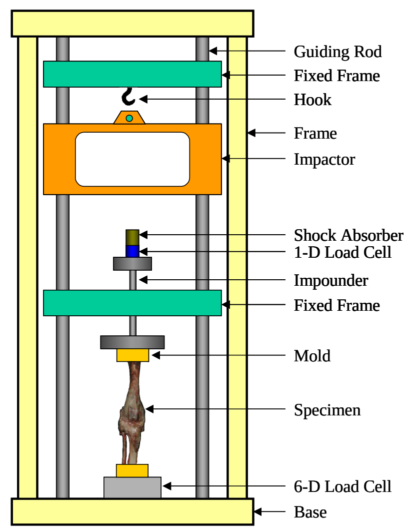
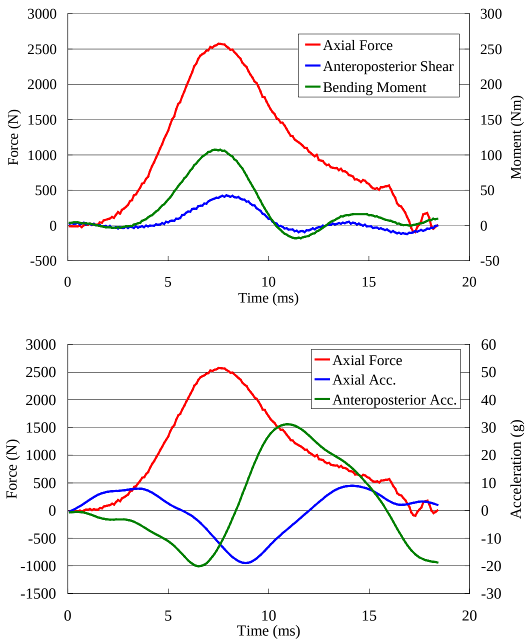


Figure 1. An impactor guided by two rods was dropped from the top of the tower to produce the impact energy. The energy was transmitted to the specimen through the impounder. The shock absorber was mounted on the top of the impounder. The knee joint specimen was placed below the impounder and above the load cell.

RESULTS

A typical loading history of forces, moment, and accelerations of knee joint is showed in Figure 2. The anterior shear force and flexion moment increased as the axial loading increased (Figure 2A). As the loading increased, the femur accelerated in the downward and anterior direction (Figure 2B). The magnitude of maximum forces, moment and axial accelerations (i.e., downward direction), and minimum anteroposterior accelerations (i.e., anterior direction) all decreased as the contact time increased. The higher energy input will results in higher mechanical responses (Figure 3). The axial force response of human and porcine joint is very close when the external energy is the same (Figure 3A). However, the shear force, moment, and accelerations of porcine joints are larger than the human one (Figure 3B,C,D,E). Most of the mechanical responses were attenuated if contact time is greater than 40 ms.



(A)

(B)

Figure 2. A typical loading history shows (A) the axial force, anteroposterior shear force and bending moment, and (B) the axial and anteroposterior accelerations. The specimen is loaded at 3.6 J impact energy using the stiffest shock absorber.

DISCUSSION AND CONCLUSIONS

We successfully developed an apparatus to find the kinetic properties of porcine and human knee joint during compressive impact loading. We found that the axial force responses of these two specimens are very similar. The deviation of the mechanical responses from human to porcine joints is higher when the contact time is short, i.e., around 20 ms; however, the deviation becomes smaller when the contact time become longer. This may indicate that the error of using porcine joint to simulate the response of human joint would be larger when the contact time is short. Few imitations should be addressed here. First, since the human knee joint came from the donation of patient after amputation, the femurs were only, in general, half or one third of the original length. This may results the smaller accelerations of human joints than the porcine specimen. Second, the specimens were fixed in both ends with no tolerance of rotations. In the real world, the ankle or the hip joints were free to move. Hence, the results of current study especially in the sagittal direction may be different from the physiological loading.

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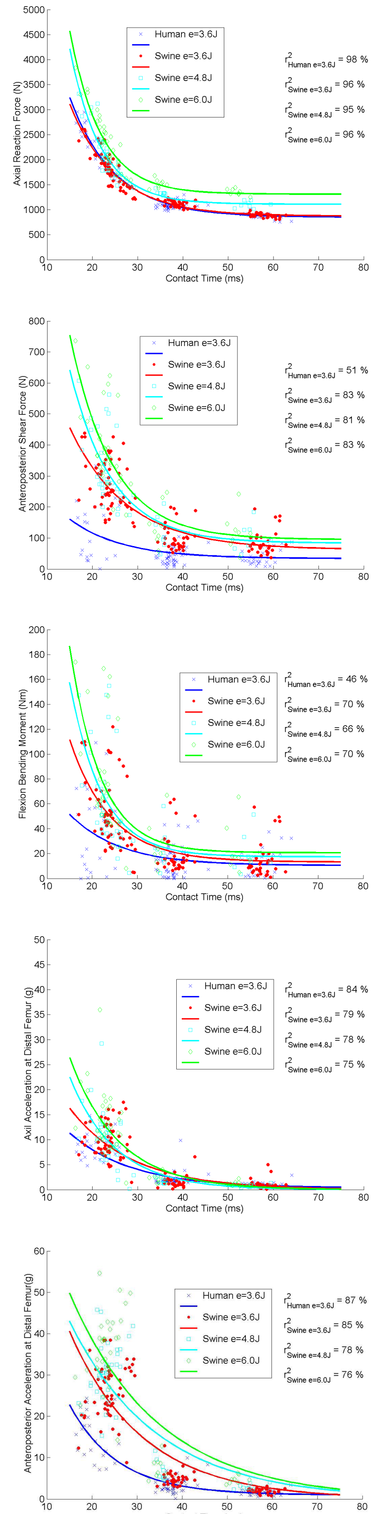
ACKNOWLEDGEMENT

We acknowledge the financial support of the national science council, roc, nsc 91-2320-b-002-159.



(A) Axial Force

(B) AP Shear Force

(C) Flexion Moment

(D) Downward Acc.

(E) Anterior Acc.

Figure 3: Mechanical responses w.r.t. contact time

THE EFFECT OF LOADING ENERGY, SHOCK ABSORBER AND KNEE JOINT POSTURES ON THE CONDYLE CONTACT PRESSURE DURING COMPRESSIVE IMPACT LOADING – AN IN VITRO BIOMECHANICAL PORCINE MODEL

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INTRODUCTION

The degenerative osteoarthritis or post-traumatic osteoarthritis is the consequence of mechanical loading on lower extremity that results in articular cartilage and bone fracture or surface fissuring [1,2]. The researches of the effect of impact loading on the knee joint includes the injury mechanics of articular cartilage and subchondral bone, the biological response of lower extremity, the kinematics and kinetics of total knee joint during impact loading. However, the dynamic property of contact pressure response of the condyle during impact loading is less studied. The objective of current study was to find the dynamic properties of contact pressure at different locations of condyle during impact loading. The relationship of contact pressure with respect to the impact energy, loading rate, knee posture including inter/external rotation and varus/valgus were also analysed.

MATERIALS AND METHODS

Eight fresh-frozen knee joints from mature swine weighted from 100 kg to 120 kg were used in the experiment. The size of porcine knee joint is close to the size of adult human knee joint. The mediolateral width and anteroposterior length of tibia plateau of eight tested porcine specimens are 72.2±2.2 and 41.27±2.2 mm, and that of human specimen are 67.1±0.2 and 41.0±0.7 mm [3]. In preparation for the testing, the specimens were stripped of all soft tissue except at the knee joint, where all ligamentous, capsular, and intracapsular structures were preserved. The specimens were than potted with quick setting epoxy at both ends. A “drop-tower type” impact testing apparatus was used for the testing (Fig 1). An impactor guided with two rods was dropped from the top of the tower to produce the impact energy. The energy was transmitted to the specimen through the impounder. The shock absorber was mounted on the top of the impounder. The shock absorber was placed on the top of the impounder to control the impact contact period. The shock absorbers are able to give, approximately, the impact contact time at levels of 20 (stiff), 40 (medium) and 60 (soft) mini-seconds. The specimen was mounted vertically below the impounder and above the six-axial force load cell (AMTI MC6-6-4000, Advanced Mechanical Technology, Inc., Watertown, MA, USA). Six thin film pressure sensors (FlexiForce Sensors, Model#A101, Tekscan Inc.) were inserted into the medial and lateral condyle (fig 2). Signals of reaction force and contact pressures were all recorded at 10 kHz sampling frequency. The signals were than low pass filtered at 1 kHz frequency using Butterworth filtering algorithm. The experimental protocol includes four factors (energy, damper, rotation, and alignment), and each factor has three levels. Every specimen was tested at 30, 40 and 50 mm levels of impact height using the three different shock absorbers. The impact weight is 12 kg; hence the impact energy input was 3.6 J, 4.8 J and 6.0 J. The posture of knee joints includes neutral, 10˚ inner and external rotation, and 3.4˚ varus and valgus. The Box-Behnken experimental design was used, hence 27 tests were done for each specimen. 216 tests in total were conduct. The loading history including the axial force, and six contact pressures were recorded via a computer. The maximum values of six contact pressures were grabbed and correlated with impact energy (E), damper (D), rotation (R) and alignment (A). The dimensions of energy and rotation/alignment are J and degree. The dimension of damper is expressed as the impact contact time that the shock absorber offered. The contact time of stiff, medium, and soft damper are 20, 40 and 60 ms. The positive rotation represents the external rotation. The positive alignment represents the varus alignment and the negative one represents the valgus. A statistical software Minitab R13 (Minitab Inc., USA) was used for the multiple variable regression analysis.

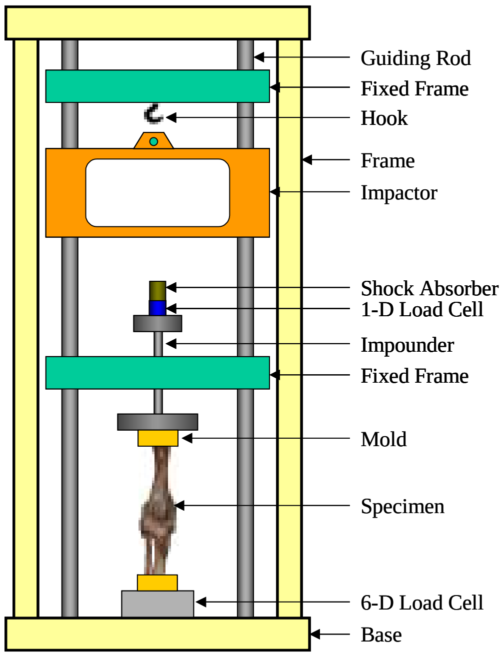

Fig 1. An impactor guided with two rods was dropped from the top of the tower to produce the impact energy. The energy was transmitted to the specimen through the impounder. The fixed frame, which is fixed to the guiding rode, is used to align the vertical movement of impounder. The knee joint specimen was placed below the impounder and above the load cell. The pressure sensors were inserted into the joint condyle.

Fig 2. The locations of six pressure sensors on the joint condyle

RESULTS

The contact pressure of condyle responded promptly with axial force (Fig 3). The contact pressures in general are larger at location 3 and 5, and are much smaller at location 2 and 6 (Fig 4). The correlation equations showed that all the peak pressure increased with impact energy, decreased with the stiffness of damper. The external rotation will increase the pressure of location 1, 2, 3 and 5, but will decrease the pressure of location 4 and 6. The varus of knee joint will increase the pressure of location 2, 3 and 5 while the valgus will increase the pressure of location 1, 4 and 6. The effect of impact energy, stiffness of damper and varus on the contact pressure is highest at location 5; while the effect of rotation on the pressure is highest at location 3 for external rotation, and at location 4 for internal rotation (Table 1).

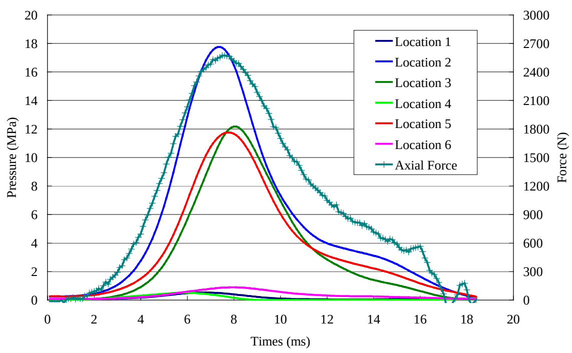

Fig 3. The example of loading history of axial force, and condyle contact pressures at six locations. The specimen is loaded at 3.6 J impact energy using the stiff shock absorber. The knee joint is in neutral posture.

DISCUSSION AND CONCLUSIONS

We successfully developed an apparatus to find the effect of loading energy, shock absorber and knee joint postures on the condyle contact pressure during compressive impact loading. The overall average of contact pressure is 3.3 MPa, which is a little bit smaller than the physiological loading condition (5-10 MPa) of human knee joint during daily activities [4]. The contact pressure is higher at locations 3 and 5 but lower at 2 and 6. It is not easy to imagine anatomically why is the higher or lower contact pressure. However, if we average 5 and 6 as lateral site, and 2 and 3 as medial site, the mean value of these four locations, i.e., medial, medial notch, lateral notch, and lateral, are quite similar. The higher or lower contact pressure may due to the intrinsic imbalance of the apparatus and may need further investigation. The external rotation of knee joint will increase contact pressure of the medial condyle and especially significant at posteromedial site, while the internal rotation will increase the pressure in most of the lateral site, and especially higher at lateral notch. The varus of knee joint will increase the pressure of medial site, and the valgus will increase the pressure of lateral notch. However, to our surprise, the varus also increased the pressure of anterolateral condyle. We do not know if this is due to the specific anatomical characteristics of porcine joint or boundary conditions of the protocol; hence need further investigation.

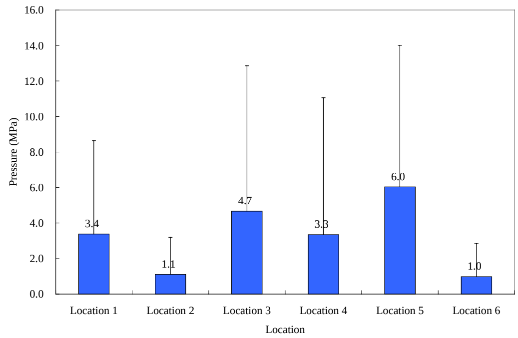

Fig 4. The mean value of peak contact pressure at six locations

Table 1. The correlation of contact pressure w.r.t. the impact energy (E, J), damper (D, ms), rotation (R, degree) and alignment (A, degree). The positive rotation represents the external rotation. The positive alignment represents the varus alignment and the negative one represents the valgus. P1 is the pressure at location 1, and P2 is of pressure at location 2, … etc.

Contact Pressure, Independent variable: E, D, R, A
P1 = 3.80 +0.447E -0.0716D +0.0336R -0.659A
P2 = 1.06 +0.213E -0.0269D +0.0495R +0.189A
P3 = 4.72 +0.763E -0.102D +0.310R +0.898A
P4 = 2.73 +0.783E -0.0850D -0.189R -1.09A
P5 = 4.47 +1.583E -0.162D +0.0021R +1.75A
P6 = 1.26 +0.086E -0.0199D -0.0327R -0.277A

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THE ACCELERATION ATTENUATION PROPERTIES OF PORCINE, INTACT HUMAN AND HUMAN KNEE JOINT AFTER MENISCECTOMY DURING COMPRESSIVE IMPACT LOADING

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INTRODUCTION

The impulse force/accelerations was produced during ambulation due to the ground reaction force. The degenerative or post-traumatic knee joint disorder is the consequence of mechanical loading on lower extremity that results in articular cartilage and bone fracture or surface fissuring [1,2]. One of the major biomechanical functions of knee joint complex is to reduce the impulse acceleration form external impact forces. The purpose of the current study is to find the acceleration attenuation of knee joint with/out meniscus using the porcine and human specimen.

MATERIALS AND METHODS

Eight fresh-frozen porcine and seven human knee joints were used. In preparation for the testing, the specimens were stripped of all soft tissue except at the knee joint, where all ligamentous, capsular, and intracapsular structures were preserved. The specimens were than potted with quick setting epoxy at both ends. A "drop-tower type" impact testing apparatus was used for the testing (Fig 1). An impactor guided with two rods was dropped from the top of the tower to produce the impact energy. The energy was transmitted to the specimen through the impounder. The shock absorber was mounted on the top of the impounder. The shock absorber was placed on the top of the impounder to control the impact contact period. The shock absorbers are able to give, approximately, the impact contact time at levels of 20 (stiff), 40 (medium) and 60 (soft) mini-seconds. The specimen was mounted vertically below the impounder and above the six-axial force load cell (AMTI MC6-6-4000, Advanced Mechanical Technology, Inc., Watertown, MA, USA). Eight two-axial accelerometers (ADXL250EB, Analog Device Inc., USA) were placed in both frontal and sagittal plan of the knee joint from top to the bottom (fig 2). Signals of reaction force and accelerations were all recorded at 10 kHz sampling frequency. The signals were than low pass filtered at 500 Hz frequency using Butterworth filtering algorithm. The experimental protocol includes three factors (damper, rotation, and alignment), and each factor has three levels. Every specimen was tested at 30 mm impact height using the three different shock absorbers. The posture of knee joints includes neutral, 10˚ inner and outer rotation, and 3.4˚ varus and valgus. The impact weight is 12 kg; hence the impact energy input was 3.6 J. The Box-Behnken experimental design was used, hence 14 tests were done for each sample. 364 tests in total were conduct. The loading histories including the axial force, and 16 channels of accelerations were recorded via a computer. Paired sample T-test was used to test the significant change in accelerations from femur distal to tibia proximal at 95% confidence level.

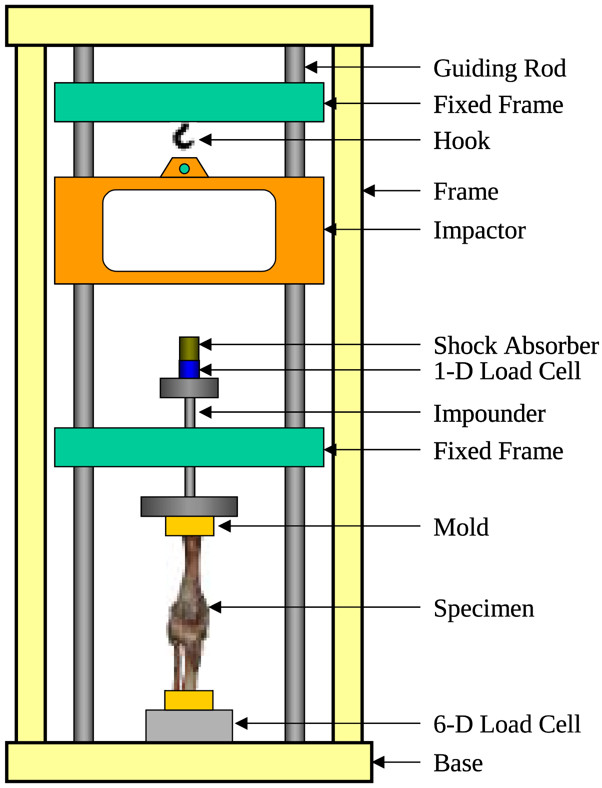


Fig 1. An impactor guided with two rods was dropped from the top of the tower to produce the impact energy. The energy was transmitted to the specimen through the impounder. The shock absorber was mounted on the top of the impounder. The fixed frame, which is fixed to the guiding rode, is used to align the vertical movement of impounder. The knee joint specimen was placed below the impounder and above the load cell.

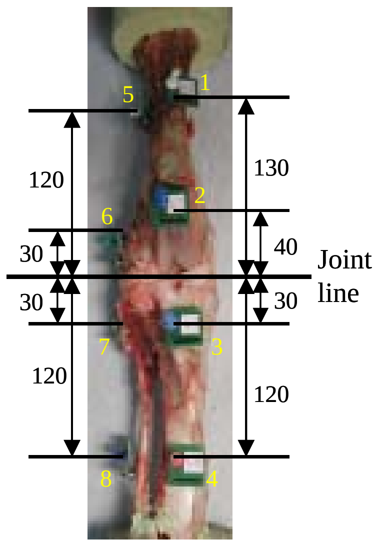
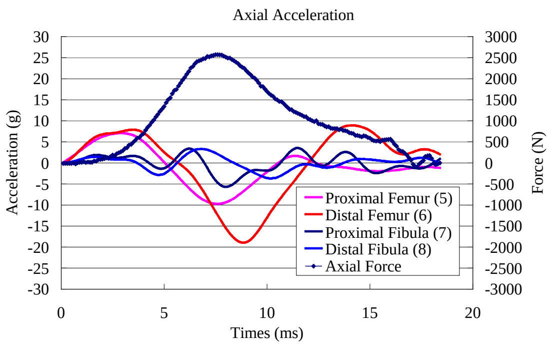


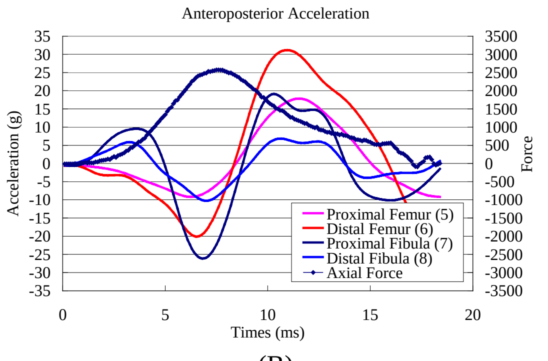
Fig 2. The locations and their dimensions (mm) of the eight accelerometers sensors on the joint condyle

RESULTS

The examples of loading history of axial force and accelerations from sensor 5, 6, 7 and 8 are showed in Fig 3. The femur site showed a larger downward acceleration than the tibia site in the beginning. As the loading increased, the femur started to rebound, i.e., the physical meaning of negative acceleration (Fig 3A). The femur accelerated in the posterior direction, and the tibia accelerated in anterior direction in the beginning of the loading. As the loading increased, all accelerations were in anterior direction. As the loading decreased, the same magnitude of posterior acceleration occurred (Fig 3B). The maximum axial accelerations (i.e., downward direction), and minimum anteroposterior accelerations (i.e., anterior direction) of three different joint were analysed and showed in Fig 4. The porcine joints obtained higher accelerations than human joints (Fig 4A,B). The joints with no meniscus showed higher anterior acceleration than intact joints, but not significant (Fig 4B). The human joint reduced higher percentages of acceleration than porcine joints in both axial and anterior direction. The joints with no meniscus reduced less percentage of accelerations than intact joints (Table 1).



(A)

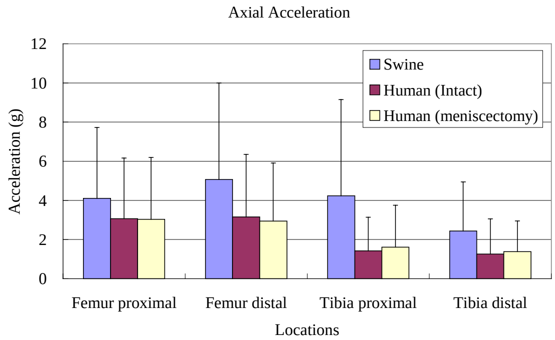


(B)

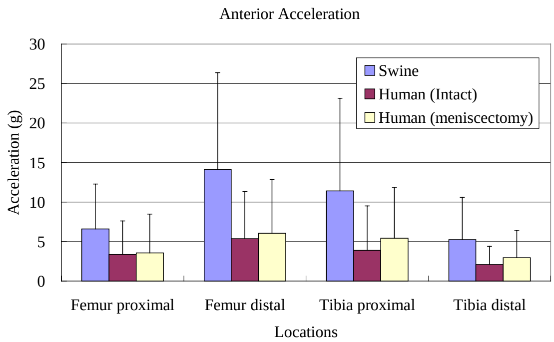
Fig 3. The examples of loading history of axial force, and (A) axial and (B) anteroposterior accelerations at locations 5, 6, 7 and 8. The specimen is loaded at 30 mm impact height, stiff shock absorber. The knee is in neutral position.

DISCUSSION AND CONCLUSIONS

We successfully developed an apparatus to find the acceleration attenuation properties of porcine, intact human and human knee joint after meniscectomy during compressive impact loading. The intact human knee joint was able to reduce 59 % of transient peak force [3], which is close to the reduction of axial acceleration found in the current study. The intact knee joint reduced 10 more percents and 20 more percents of axial and anterior accelerations than the meniscectomic one. This may indicate that the meniscus contributes 10 and 20 percents of axial and anterior acceleration reduction. Few imitations should be addressed here. First, since the human knee joint came from the donation of patient after amputation, the femurs were only, in general, half or one third of the original length. This may results the smaller accelerations of human joints than the porcine specimen. Second, the specimens were fixed in both ends with no tolerance of rotations. In the real world, the ankle or the hip joints were free to move. Hence, the results of current study especially in the sagittal direction may be very different from the physiological loading.



(A)



(B)

Fig 4. The (A) axial and (B) anterior accelerations of knee joint at four different locations.

Table 1. The reduced percentage from femur distal to tibia proximal at two directions of accelerations

	Axial	Anterior
Swine	16 % (p=0.037)	19 % (p<0.001)
Human (intact)	55 % (p<0.001)	28 % (p=0.002)
Human (meniscectomy)	45 % (p<0.001)	10 % (p=0.194)

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THE SHOCK ATTENUATION PROPERTIES OF STRAIGHT STANDING KNEE JOINTS USING DIFFERENT SHOCK ABSORBERS AND ENERGY INPUTS

Jaw-Lin Wang* and Yen-Lin Lee

ABSTRACT

Exposure of human lower extremities to mechanical loading will lead to different biological responses, e.g., degenerative and post-traumatic osteoarthritis. Relations among injuries and impact energy have been well studied. However, how the external force is attenuated by the internal joint is not known yet. The objective of the current study is to find the shock attenuation properties of knee joint using different shock absorbers. A “drop-tower type” impact apparatus was used for testing. Ten fresh porcine knee joints were aligned at full extension for testing. All specimens were tested at 30 and 40 mm levels of height, 12 and 16 kg levels of weight, and with shock absorbers of 20 (stiff), 40 (medium), and 60 (soft) mini-seconds levels of contact time. The impact forces, flexion moment and all directions of acceleration were found to decrease with a softer shock absorber. A medium shock absorber produces 33% less axial force, 68% less shear force, 53% less bending moment and more than 70% less accelerations at femur and tibia compared to the stiff shock absorber. The results of the current research can be used for reference in designing sports footwear.

Key Words: knee joint, impact biomechanics, shock attenuation.

I. INTRODUCTION

Exposure of the human lower extremities to severe mechanical loading will lead to different biological responses. Degenerative osteoarthritis is induced by a high magnitude and rate of repetitive impact loading during locomotion (Radin *et al*., 1982; Radin *et al*., 1972; Simon *et al*., 1972); while post-traumatic osteoarthritis is the consequence of blunt impact force applied to articular cartilage that results in bone fracture or surface fissuring (Oegema *et al*., 1993; Thompson *et al*., 1991; Vener *et al*., 1992). The loading, in the former case, may be the result of the sports activities, while the blunt impact may be the result of a traffic accident.

Research on the effect of impact loading on the knee joint includes only a few major categories; e.g., local injury mechanics (Radin *et al*., 1973; Repo and Finlay, 1977; Torzilli *et al*., 1999; Zhang *et al*., 1999) and biomechanics (Radin and Paul, 1970; Radin *et al*., 1970) of components, such as articular cartilage or subchondral bone, the biological response of lower extremity due to blunt impact loading (Atkinson *et al*., 2001; Atkinson and Haut, 1995; Atkinson and Haut, 2001a; Atkinson and Haut, 2001b; Donohue *et al*., 1983; Hurwitz, 1984; Thompson *et al*., 1991), and global biomechanics such as kinematics and kinetics of total knee joint during impact loading (Chu *et al*., 1986; Hoshino and Wallace, 1987). Although there are many knee injury studies, how external devices or strategies can help to minimize knee injury from external impact loading has still not been addressed.

A durable cushioning system in the shoe is proved to be necessary for the protection of the knee joint from sports activities such as distance running and jumping. A modified basketball shoe with improved shock attenuation significantly reduced the incidence of metatarsal stress fractures and foot overuse injuries resulting from vertical impact loads, compared with standard infantry boots (Milgrom *et al*.,

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Table 1 The dimensions (mm) of mediolateral width and anteroposterior length of tibia plateau of eight tested porcine specimens in comparison with that of a human one

Dimension of Tibia Plateau (mm)	Mediolateral Width (ML)	Anteroposterior Length (AP)
Swine, n=8 (Current study)	72.2±2.21	41.27±2.2
Human, n=16 (Cheng *et al*., 1999)	67.1±0.19	41.0±0.72

1992). Nevertheless, there is no quantitative information on how soft the footwear should be to prevent the injury, and the reason of why a soft shoe heel would protect the knee has not been given.

The objective of the current study was to find the shock attenuation properties of knee joints using different shock absorbers to mimic different stiffnesses of shoe heel. The kinetic information such as axial force, anteroposterior shear force and bending moment, as well as kinematic information such as accelerations of femur and tibia both in lateral and sagittal directions were recorded for analysis to find the quantitative attenuation properties of straight standing knee joints.

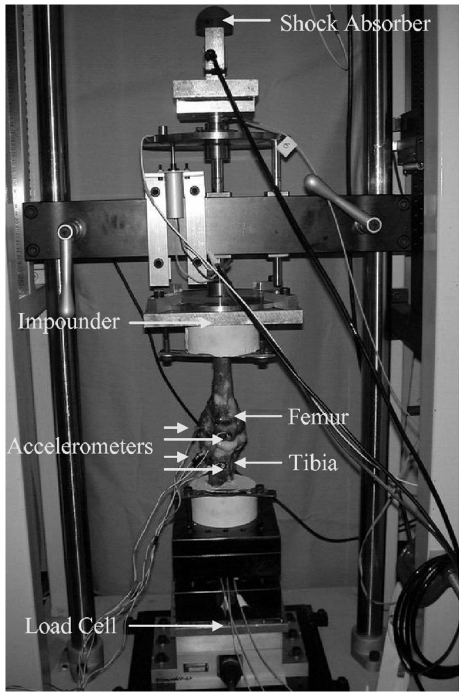

Fig. 1 The porcine knee joint was mounted in the drop-tower type impact testing apparatus. An impactor (not shown in the figure) guided by two rods was dropped from the top of the tower. The shock absorber was mounted on the top of the impounder. The knee joint was placed below the impounder and above the load cell. The six-dimensional force responses and accelerations of femur and tibia in sagittal and lateral directions were recorded during impact

II. MATERIAL AND METHOD

1. Specimen Preparation

Ten fresh-frozen knee joints from mature swine, which weighed from 100 kg to 120 kg were used in the experiment. The dimensions of important anatomical characteristics of knee joint are close to the size of human adult knee dimensions (Cheng *et al*., 1999) (Table 1). The bone mineral density (BMD) of 92 kg and 120 kg porcine back legs ranges from 1.053 to 1.183 g/cm^2 (Mitchell *et al*., 2001), which is close to the quality of distal femurs of 14 to 16 year old Caucasian girls (Henderson *et al*., 2002). In preparation for the testing, the specimens were stripped of all soft tissue except at the knee joint, where all ligamentous, capsular, and intracapsular structures were preserved. For better alignment, the joint complexes were fixed with external fixators first. The specimens were than potted with quick setting epoxy (Polyesterputty, Gou-Fon, Inc., Taipei, Taiwan) at both ends. Four accelerometers (PCB 353B17, PCB Piezotronics, Inc., Depew, NY, USA) were mounted at the anterior and medial sides of the femur and the tibia. The measuring axis of the accelerometers was perpendicular to the mounted bone surface. Hence, the acceleration of femur and tibia in both sagittal and lateral directions can be measured.

2. Experimental Apparatus

A “drop-tower type” impact testing apparatus was used for the testing (Fig. 1). An impactor (not shown in the figure) guided by two rods was dropped from the top of the tower. The shock absorber was placed on the top of the impounder to control the impact contact period. The designed stiff, medium, and soft shock absorbers made of plastic rubber and silicon rubber, are able to give, approximately, contact times of impact between impounder and impactor at levels of 20, 40 and 60 mini-seconds when testing a rigid still bar specimen at 12 kg impact mass and 50 mm impact height (Lee, 2002). The specimen was mounted vertically below the impounder and above the six-axial force load cell (AMTI MC6-6-4000, Advanced Mechanical Technology, Inc., Watertown, MA, USA). Signals of three dimensional reaction forces and moments from the six-axial force load cell and accelerations of tibia and femur were all recorded at 10 kHz sampling frequency. The signals were than low pass filtered at 1 kHz frequency using Butterworth filtering algorithm.

3. Experimental Protocol

All specimens were tested at 30 and 40 mm levels of height, and at 12 and 16 kg levels of weight, using three different shock absorbers. The energy input was then calculated at the levels of 3.6 J, 4.8 J and 6.4 J. A full factorial experimental design was used; hence, each specimen was tested 12 times and 120 experiments in total were conducted. Ten minutes rest, at least, was given between impacts. The total experimental time for one specimen was finished in four hours to minimize the chronicle variation of mechanical strength of knee joint. The loading history including the axial reaction force, anteroposterior shear force, flexion bending moment and accelerations of femur and tibia at sagittal and lateral planes were recorded via a computer. The maximum values of the above dependent variables were later analyzed.

4. Statistical Method

We divided the maximum values of dependent variables into three levels of input energies (3.6 J, 4.8 J, 6.4 J), and then compared the effect of stiff, medium and soft shock absorbers at each level of energy input. The over all responses of dependent variables with respect to the shock absorbers were also compared to give a global idea of how the shock absorber affected knee mechanics. One-way analysis of variance (ANOVA) was used to test the significant change in forces and accelerations from the stiff to medium shock absorber, and from the medium to soft shock absorber at the 95% confidence level. A paired sample T-test was conducted to compare the variations between two dependent variables, for example, the differences of accelerations from sagittal to lateral directions, or differences from femur to tibia, ... etc.

III. RESULTS

1. A Typical Impact Pattern

All specimens remained intact, i.e., no observable bony or cartilage fracture was noted through out the experiment by visual examination. A typical force, bending moment and acceleration loading history of neutral standing knee joint during impact loading using three different shock absorbers is plotted in Fig. 2 (Specimen #1, Impact Mass = 12 kg, Impact Height = 30 mm). The axial reaction force shows a nearly symmetrical haversine, while the anteroposterior shear force shows a full sine-wave curve within one impact period. The bending moment was in flexion mode during most of the loading, but with a minor extension at the beginning and the end of the loading. The accelerations of tibia and femur in both sagittal and lateral directions responded promptly with the shear force. The shear force and accelerations are in the posterior direction first, and than reversed to the anterior direction at about 7 miniseconds after impact.

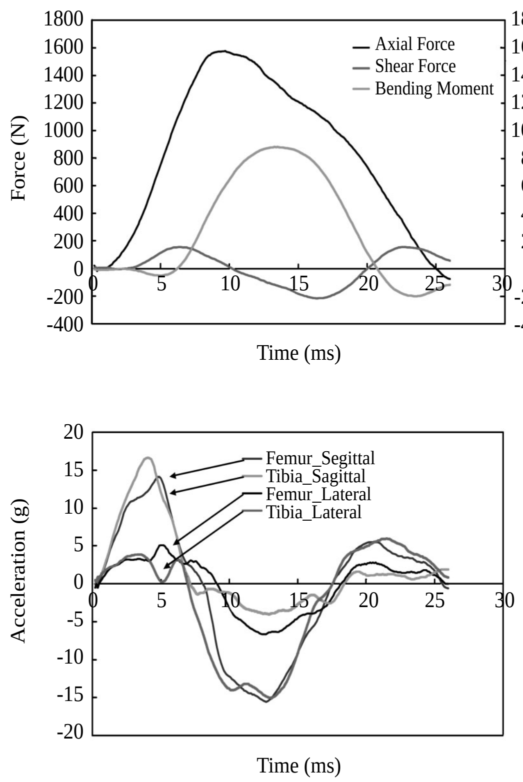

Fig. 2 The typical loading history (Specimen #1) of (a) the axial reaction force, anteroposterior shear force, and flexion bending moment, and (b) the accelerations of femur and tibia in sagittal and lateral directions using the stiff shock absorber. The specimen is loaded at 12 kg impact mass and 30 mm impact height

2. Relations between Force Responses and Energy Input

Higher energy input will increase the responses of forces, moments and accelerations. The averaged axial reaction force, anteroposterior shear force, and flexion moment reach 1904 N, 221 N and 67 Nm when using the stiff shock absorber at 3.6 J energy input. Comparing the values of forces and moment at 3.6 J to 6.4 J energy input. They increased 78%, when using the stiff shock absorber, the peak axial force, shear force and bending moment increased 28% (p=0.125),

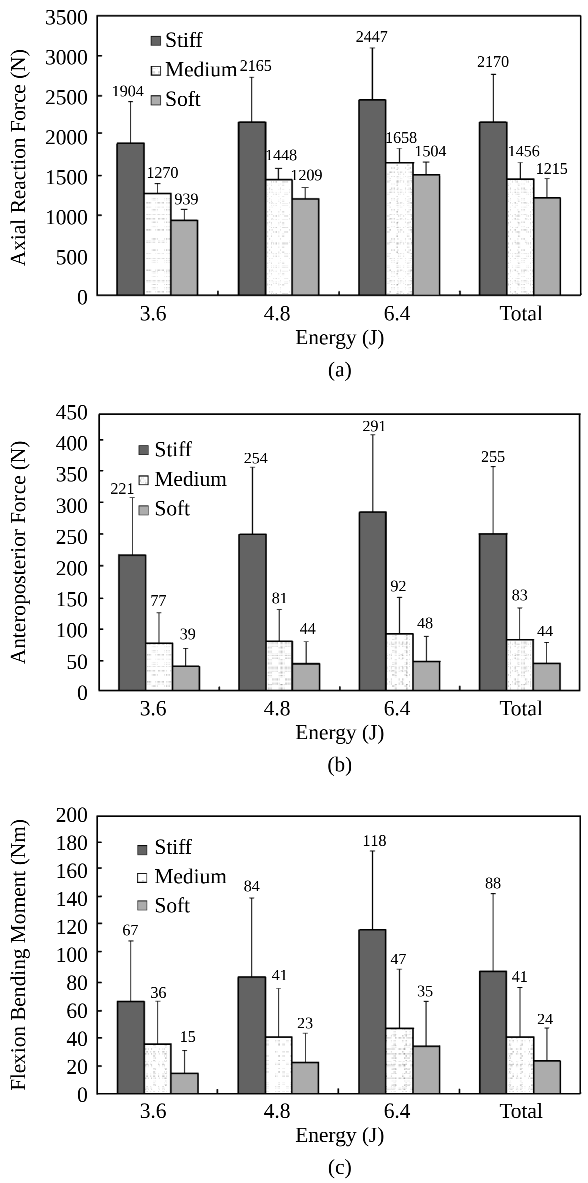

Fig. 3 (a) The axial reaction force; (b) the anteroposterior shear force, and (c) the flexion bending moment at three levels of energy input and three different shock absorbers

32% (p=0.371), and 76% (p=0.115), respectively (Fig. 3). The increased percentage of bending moment is close to the increased percentages of energy input.

3. Acceleration Attenuation

The acceleration in the sagittal direction is significantly higher than the acceleration in the lateral direction (35g vs. 16g at femur (p<0.001) and 27g vs. 13g at tibia (p<0.001)) when using the stiff shock absorber. The medium shock absorber will reduce the most accelerations in all directions and registers below 10g, while the soft shock absorber will reduce

Table 2 The average (± standard deviation) of maximum value of the axial reaction force (F_{Axial}), anteroposterior shear force (F_{Ant}), flexion bending moment (M_{Flex}) and accelerations of femur and tibia in sagittal and lateral planes ($G_{FemurSag}$, $G_{TibiaSag}$, $G_{FemurLat}$, $G_{TibiaLat}$) at two impact masses when the impact energy is fixed at 4.8 J. One-way analysis of variance (ANOVA) was tested to determine the differences between the 12 kg impact mass and 16 kg impact mass

	Mass=12 kg	Mass=16 kg	p-value
F_{Axial} (N)	1577±552	1637±526	0.669
F_{Ant} (N)	131±122	122±113	0.762
M_{Flex} (Nm)	50±49	48±47	0.875
$G_{FemurSag}$ (g)	16±15	15±18	0.811
$G_{TibiaSag}$ (g)	13±13	11±11	0.498
$G_{FemurLat}$ (g)	7.0±6.9	6.6±6.1	0.840
$G_{TibiaLat}$ (g)	6.7±7.2	5.8±5.9	0.582

the accelerations to below 3g. The knee joint itself reduced 32% (p=0.039), 22% (p=0.012) and 13% (p=0.108) of sagittal accelerations from femur to tibia at energy levels of 3.6 J, 4.8 J and 6.4 J when using the stiff shock absorber, while the lateral accelerations reduced -5% (p=0.753), 7% (p=0.620) and 38% (p=0.220), respectively. The average reductions of accelerations of knee joint are 23% (p<0.001) in sagittal direction and 19% (p=0.210) in lateral direction while using the stiff shock absorber (Fig. 4).

4. Effect of Impact Mass when the Impact Energy Is Fixed

When the impact energy is fixed at 4.8 J, we have two combinations of energy input, i.e., 12 kg impact mass with 40 mm impact height, and 16 kg impact mass with 30 mm impact height. We compared the results of force, moment and acceleration responses with respect to two impact masses, and found no significant difference between these two groups (Table 2).

5. Effect of Shock Absorber

The soft shock absorber decreased the forces and accelerations. The over all responses (i.e., including all three levels of input energy) of axial force, shear force and flexion moment significantly decrease 33% (p<0.001), 68% (p<0.001) and 53% (p<0.001) from stiff to medium shock absorber, respectively. The medium shock absorber reduces more than 70% of

Table 3 The average (± standard deviation) of maximum value of the axial reaction force (F_{Axial}), anteroposterior shear force (F_{Ant}), flexion bending moment (M_{Flex}) and accelerations of femur and tibia in sagittal and lateral planes ($G_{FemurSag}$, $G_{TibiaSag}$, $G_{FemurLat}$, $G_{TibiaLat}$) at three different stiffness levels of shock absorbers, and the decreased percentage of dependent variables from stiff to medium shock absorber, and from medium to soft shock absorber. One-way analysis of variance (ANOVA) was tested to determine the differences between the stiff and medium shock absorber, and between the medium and soft shock absorber

Shock Absorber	Stiff	Medium	Soft	Stiff-Medium $(V_{Stiff} - V_{Medium})/V_{Stiff}$	Medium-Soft $(V_{Medium} - V_{Soft})/V_{Stiff}$
F_{Axial} (N)	2170±597	1456±207	1215±247	33%, p<0.001	11%, p<0.001
F_{Ant} (N)	255.0±110.0	82.7±51.8	44.0±34.5	68%, p<0.001	15%, p<0.001
M_{Flex} (Nm)	88.3±56.0	41.3±35.3	23.7±23.7	53%, p<0.001	20%, p=0.011
$G_{FemurSag}$ (g)	34.7±13.2	9.4±6.6	3.3±2.6	73%, p<0.001	17%, p<0.001
$G_{TibiaSag}$ (g)	27.0±9.4	6.2±3.0	2.6±1.3	77%, p<0.001	14%, p<0.001
$G_{FemurLat}$ (g)	15.8±12.3	3.9±3.3	2.3±1.1	75%, p<0.001	11%, p=0.004
$G_{TibiaLat}$ (g)	13.2±7.2	3.4±1.3	2.1±1.1	74%, p<0.001	10%, p<0.001

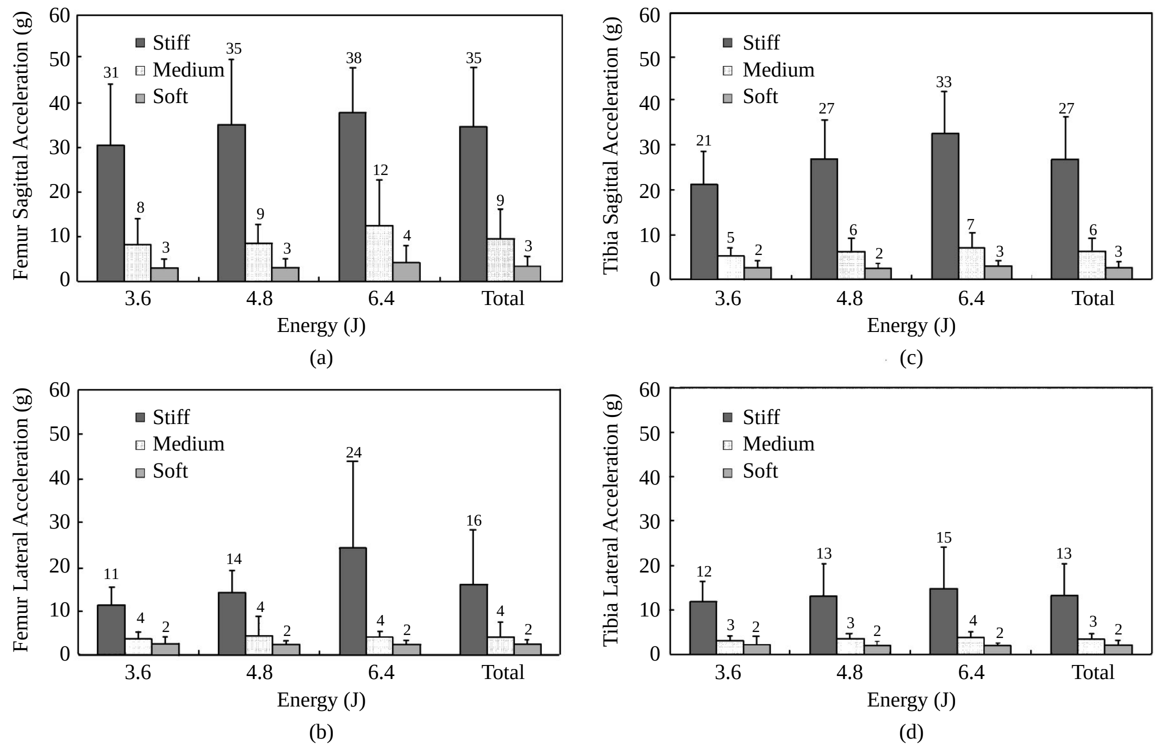

Fig. 4 Accelerations at (a) femur sagittal; (b) femur lateral; (c) tibia sagittal, and (d) tibia lateral directions of three different energy inputs and three different shock absorbers

shock accelerations on femur and tibia on lateral and sagittal planes compared to the stiff shock absorber. The soft shock absorber further absorbs force and acceleration; however, all decreases are under 20%, and not every variation is significant (Table 3). It is concluded that the medium shock absorber, i.e., shock absorber with 40 mini-seconds of contact time, has greatly and efficiently reduced the forces and accelerations around the knee joint during impact.

IV. DISCUSSION

We successfully developed a testing apparatus to find the six-dimensional mechanical force response of complex structures during impact loading. The major findings of the current study are; first, the increased percentages of flexion bending moment are well predicted by the increase of energy input; second, the knee joint is capable of reducing by 23% sagittal

Table 4 The magnitude of energy input of knee joint impact studies in literature

Researcher	Energy (J)	Species	Joint	Method
(Lafortune *et al*., 1996b)	36.25	Human	Tibiofemoral	In vivo
(Hoshino and Wallace, 1987)	17.6	Human	Tibiofemoral	In vitro
(Atkinson *et al*., 2001)	11	Canine	Tibiofemoral	In vitro
(Fukuda *et al*., 2000)	0.46	Swine	Tibiofemoral	In vitro
(Thompson *et al*., 1991)	66	Canine	Patellofemoral	In vivo
(Donohue *et al*., 1983)	13.7	Canine	Patellofemoral	In vivo
(Haut *et al*., 1995)	6.3	Rabbit	Patellofemoral	In vivo
(Newberry *et al*., 1997)	6.0	Rabbit	Patellofemoral	In vivo

accelerations; and third, the medium shock absorber can efficiently reduce 30% of axial force and 70% of accelerations over the stiff shock absorber.

The current study used porcine knee joints to mimic human knee joints. The size of the porcine knee joint is very close to the human adult knee joint's size. However, the full extension posture of the porcine knee joint is slightly flexed (5 to 10 degrees), compared to the posture of the human joint. This may indicate that the human knee joint may experience higher axial force, but lower anteroposterior shear force and flexion moment at the same loading condition. The in vitro study neglects the damping effect provided by the muscles and heel fat pad. The results of the current study may represent the worst scenario of impact phenomena on knee joints during impact loading.

The energy input of the current study ranges from 3.6 J to 6.4 J, which is lower than the protocols of numerous knee joint impact studies (Table 4). The average of maximum axial force reaches 2,450 N at 6.4 J energy input using the stiff shock absorber. Assuming the contact area of a swine knee joint is about 1000 mm^2 (Fukubayashi and Kurosawa, 1980; Ihn *et al*., 1993), the stress at the cartilage surface will achieve 2.5 MPa. In the literature, the peak ground reaction force (GRF) will reach 2 to 10 times body weight during running and jumping actions (Cavanagh and Lafortune, 1980), and the estimated physiological contact stress of a weight bearing joint is from 5 to 10 MPa (Hodge *et al*., 1989; Kotzar *et al*., 1991). The produced contact stress of the current study is lower than the physiological contact stress. The reason for designing for low contact stress and low energy input is that the current study focused on the shock attenuation and kinematical response of knee joints with respect to different shock absorbers and energy inputs, but not the traumatic outcomes due to impact loading.

Lafortune et al. tested 21 human subjects and found that the softer shock absorber between the foot and floor is more likely to protect the lower limb system against impact loading than knee angle strategies (Lafortune *et al*., 1996a; Lafortune *et al*., 1996b). In this study, we showed the soft shock absorber already greatly reduce the contact force and acceleration around the knee joint. For a situation where the knee strategy is not an option, for example, running and jogging, the only way to reduce the contact stress is to find a shock absorber of proper stiffness.

The reason why the increase of flexion bending moment corresponds to the energy input, but not to the axial force or shear force is not clear. It may be that the physical dimension of energy, i.e., J, is equivalent to bending moment, i.e., Nm. The injury criterion of knee joint complex is not well defined, either. The most frequently used controlling variable is the energy input. However, in a study of the influence of impact energy and impact mass on the patellofemoral joint (Atkinson *et al*., 2001), the results showed higher impact mass will produce injury at all levels of impact energy. In our studies, we found the impact mass, nevertheless, does not significantly affect the magnitude of forces, moment or acceleration when the energy input is fixed. Since the variation of impact mass in our protocol was only 4 kg, i.e., 25%, a more extensive experimental design may be needed for extreme cases.

The force transmission and the shock attenuation properties of the knee joint itself during impact loading were not focused on the current study. However, we found the significant reduction of sagittal acceleration from femur to tibia to be 23%. Although this may not directly correspond to our results, Voloshin also found the shock absorbing capability, i.e., the reducing of axial acceleration in this research, of a knee joint to be less by 20% after removal of meniscus (Voloshin and Wosk, 1983). The shock absorbing function from knee joint to shoe pad may not be easy to discriminate only by the magnitude of forces or accelerations. A more detailed dynamic mathematical model may be needed to simulate the impact responses of the lower extremity complex including knee joint and shoe pad.

In our study, the results showed the low stiffness shock absorber reached high shock absorption around the knee joints. A shock absorber that is capable of providing 40 mini-seconds of contact time will greatly reduce the forces and accelerations on the joint. However, lower stiffness of shock absorber means higher rate of energy utilization during running (McMahon *et al*., 1987). Finding the perfect stiffness of shock absorber to prevent serious knee injury while maintaining energy efficiency is a question requiring further investigation. The results of this study may be used as reference for designing sports footwear.

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