Test Condition Optimization for Testing of Hip Protectors

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Abstract. Determining the efficacy of hip protectors designed to reduce the menace of the alarming rise in hip fracture due to fall among the elderly population is particularly challenging. Hip protector evaluation has been done using different soft tissue surrogate and at different energy level to determine the effectiveness of various hip protectors. However, some hip protectors that had been ranked effective in mechanical testing systems proof otherwise when deployed for clinical trials, this may be due to lack of optimization of the testing condition, most notably in testing when the soft tissue surrogate attenuates impacts that may be unduly attributed to the effectiveness of the hip protector. In this study, the performance of a surrogate soft tissue, fabricated from polyethylene, was evaluated at different energy levels (3.68 J – 37.9 J) to ascertain the optimal position for evaluating the efficacy of a hip protector to simulate a representative condition of an actual fall to the sideways by a person. A drop-weight impact testing machine was used to evaluate the impact force response of an employed femoral geometry with and without the soft tissue at various residual impact energy. The result showed that the soft tissue might be responsible for up to about 95.17% of peak impact force attenuation if not adequately modelled. Therefore various conditions were examined to get representative sideways impact condition where the soft tissue employed do not attenuate beyond 11 - 28% of the peak impact force. This finding demonstrates that the impact condition is as critical as all other parameters in determining the performance of a hip protector.

Keywords: Test system, soft tissue, hip protectors, fall and impact force.

1. Introduction
Epidemiological studies have estimated that there would be an exponential increase in the incidence of osteoporotic fractures in Asia so that by 2050, 50% of all hip fractures would occur in this region [1]. When falls cannot be prevented, and the coordinative response of the body is too slow to break the fall, more force will be applied to the hip, making it susceptible to fracture most especially in osteoporotic bone [2]. Hip protectors prove to be an effective strategy in reducing the impact force to the hip in a sideways fall [3]. Biomechanical test and clinical trials have proven that hip fracture can be drastically reduced by wearing a hip protector [4 - 8]. Researchers have, therefore, designed various types of hip protectors to prevent hip fracture in a sideways fall [8]. Lauritzen et al. [9] showed that hip fractures in the elderly could be reduced by 50 per cent if they wore a particular design of protector.

By using mechanical testing systems, researchers have been able to provide significant insights into the efficacy of a hip protector in the prevention of hip fracture [10]. However, test results are often
affected by the test conditions, mainly when tests are not carried out at representative condition of fracture causing impact [11]. Current direction is to ensure that only clinically effective hip protectors are deployed to the market. Therefore, the protective value of a hip protector must be adequately measured with an optimized test system. The influence of soft tissue is often time overlooked and not adequately modelled in many research works. Thinner soft tissue would allow higher forces to be exerted on the hip compared to a thicker one [12,13] and inappropriately chosen impact mass and residual impact energy would adversely affect results of experiments to evaluate the efficacies of hip protectors [14]. Therefore, it is appropriate to ensure that the hip assembly response as close as possible to the real situation in the test condition to evaluate the effectiveness of a hip protector.

Hence, this research aims to systematically establish a baseline force, representative of fracture causing situation, before measuring the force attenuation capabilities of hip protectors. Since the soft tissue also attenuates some impact force, the optimum test condition should be between peak force, not less than 2500 N [15] and the impact attenuation ability of the soft tissue around $16 \pm 7.8$ (SD) % [16].

2. Methodology

2.1. Average hip surface geometry
The surface geometry of the hip was taken from the three-dimensional (3D) coordinates describing the average surface geometry of the hip, buttock and anterior thigh region of older women in a previous study where participants consisted of 15 Canadian women with a mean age of 77.5 years, mean body mass of 61.2 kg, mean height of 1.61 m, and mean body mass index of 23.6 kg/m$^2$ [11]. The 3D coordinates were plotted in SolidWorks 2016 (Figure 1) to create the surface geometry.

![Figure 1](image1.png)

**Figure 1.** (a) Side view of hip surface geometry, (b) Top view of hip surface geometry used in creating the soft tissue.

The surrogate soft tissue was made using closed-cell polyethylene foams. A closed-cell polyethylene foams LD45 of density 45 kg/m$^3$ was used directly over the proximal femur and closed-cell copolymer foam Evazote EV50 of density 50 kg/m$^3$ was used over the regions anterior, posterior, and superior to the femur. The LD45 and EV50 polyethylene foam were glued together using a spray adhesive (Scotch Super 77) to form a single (220 × 150 × 90) mm$^3$ block. The block was then placed into a 3-axis CNC
machine to machine the surface geometry of the human hip according to the SolidWorks model. The thickness of the foam above the greater trochanter was set to be 2.4 cm thick [11] (Figure 2).

![Figure 2.](image)

2.2. Simplified Femur
Based on the geometry of femur employed by Keenan and Evans [15], a simplified model of the femur having the profile of the greater trochanter as traced from a Sawbones 4th generation computer-aided design (CAD) model in SolidWorks 2016 (Figure 3) was adopted for the testing. The Aluminium block (195 x 40 x 50) mm was machined by using a 3-axis computer numerical control (CNC) milling machine (Robodrill α-T21iFLb).

![Figure 3.](image)

2.3. Experiment with impact simulator
A test rig that closely adhered to the recommendation of the consensus from the International Hip Protector Research Group for biomechanical testing at the Copenhagen Consensus Conference [17] on the design of hip protector test rig was designed (Figure 4) but with the adoption of lower effective mass following the findings of researchers such as Mills [18] and Kannus et al. [5] and with no spring [19–22] having the intent of producing forces above the reported fracture threshold of 3.472 kN and observing the condition that mimics the impact attenuation provided by the human soft tissue [23,24].
Since it has been demonstrated that the simulated impact condition in a fall can be achieved by various combination of weights and heights [11], masses of 3.75 Kg, 4.75 kg and 5.52 Kg were dropped from a height of 100 mm, 300 mm, 500 mm, 600 mm, 700 mm, and 750 mm with 0.99 m/s, 1.72 m/s, 2.21 m/s, 2.43 m/s, 2.62 m/s and 2.71 m/s velocity respectively to ascertain the suitable drop impact height and femoral neck force for the hip arrangement to produce a representative fracture-causing impact in a surrogate hip arrangement containing an isolated femur. Hence, the impact energies were about 3.68 J, 11.04 J, 23.3 J, 27.96 J, 32.62 J, and 34.95 J, respectively. All tests were repeated three times for each impact condition.

The force transmitted to the bare femur was first ascertained at the femoral neck, followed by the force attenuated by the surrogate soft tissue. The force at the femoral neck when the soft tissue was not in place is the unprotected force \( F_u \) used in calculation and \( F_p \) is the impact recorded at the same point when the polyethylene foam protected the femur geometry. The percentage impact attenuation rate is defined as follows [25].

\[
\text{Impact Attenuation rate} \,(\%) = \left( 1 - \frac{F_p}{F_u} \right) \times 100\%
\]  

(1)

Figure 4. Graphical representation of the drop impact tester. In this case, simply justify the caption so that it is as the same width as the graphic.

3. Results
Considerable disparities were identified between the force attenuation provided by the soft tissue at different residual impact energy. The peak force was reduced up to 95.17% by the soft tissue when the impact weight was very low and up to 11.92% at very high impact energy. These differences in impact force attenuation demonstrate that the impact attenuation of the trochanteric soft tissue has to be correctly modelled when designing a compliant test system. The mean maximum peak force attenuation
that is representative of impact force on the femoral neck in a sideways fall were observed at a residual impact energy of 32.5 J – 37.9 J that corresponds to 2.62 m/s – 3.71 m/s velocity of impact as shown in Table 1. These velocities correspond to the reported values of velocity in a typical fall [11,12]. The peak impact force attenuated is very high when lower impact mass and height of dropping the impact mass were low. Hence, these two factors were considerably increased one factor at a time to reach a representative test condition with minimal runs of the experiment.

Table 1. Drop Impact test results for polyethylene foam and aluminium femur

| Height (cm) | Weight (kg) | Velocity (m/s) | Impact Energy (J) | Unprotected Force (kN) | Soft tissue protected force (kN) | Attenuation rate (%) |
|------------|-------------|----------------|-------------------|------------------------|-------------------------------|---------------------|
| 10         | 3.75        | 1.40           | 3.68              | 3.52 ± 0.00           | 0.17 ± 0.00                   | 95.17%              |
| 30         | 3.75        | 2.43           | 11.04             | 3.56 ± 0.00           | 0.22 ± 0.00                   | 93.82%              |
| 50         | 4.75        | 3.13           | 23.30             | 3.86 ± 0.01           | 2.00 ± 0.02                   | 48.19%              |
| 60         | 4.75        | 3.43           | 27.96             | 6.40 ± 0.00           | 4.10 ± 0.00                   | 35.94%              |
| 70         | 4.75        | 3.71           | 32.62             | 6.12 ± 0.03           | 4.54 ± 01                    | 25.82%              |
| 75         | 4.75        | 3.84           | 34.95             | 6.88 ± 0.02           | 6.06 ± 0.00                   | 11.92%              |
| 40         | 5.52        | 2.80           | 21.66             | 10.00 ± 08            | 2.53 ± 01                    | 74.66%              |
| 50         | 5.52        | 3.13           | 27.08             | 11.52 ± 02            | 4.84 ± 00                    | 57.99%              |
| 60         | 5.52        | 3.43           | 32.49             | 12.02 ± 02            | 7.01 ± 01                    | 41.65%              |
| 70         | 5.52        | 3.71           | 37.91             | 12.85 ± 02            | 9.29 ± 08                    | 27.74%              |

From the experiment conducted, the optimized arrangement for carrying out the impact attenuation test of a hip protector using our test system is gotten at an impact load of 4.75 kg at the height of 0.6 m - 0.7 m that produces an impact force within the range of 4.54 to 6.06 kN or with an impact mass of 5.52 kg at the height of 0.7 m for a more severe fall up to about 9.29 kN. At this height, the average force attenuated by the foam hip model and the aluminium femur is between 11% -27.74% which is close to the amount of force attenuated by the human tissue as previously reported [12,24,26]. Though other impact masses/impact heights variation can be employed [11] depending on the stiffness and other characteristics of the surrogate soft tissue employed, the force attenuation of the soft tissue used must first be ascertained to avoid the possibility of misrepresentation of the capability of a hip protector. Figure 5 shows the soft tissue reduces the impact force on the trochanter and increases the time to peak impact force. The experiment demonstrates the impact attenuation provided by the soft tissue could also influence hip protector testing result at different energy levels for falls to the sagittal plane.

This experiment reveals that the impact force in a sideways’ fall and the percentage of attenuated force representative of an actual fall situation in human when the sagittal plane can be achieved despite the lower impact energy employed in this study when compared to previous results [8,16].
Figure 5. A Typical femoral neck reaction force versus time for impacts to the hip assembly with and without the soft tissue sample.

4. Conclusion
The conducted experiment shows that preliminary optimization of the test condition would ensure unrealistic impact energy or too severe femoral neck force will not be employed in evaluating the performance of a hip protector. The potential of overestimating or underestimating the capability of a hip protector is therefore eliminated, thereby ensuring only biomechanically effective hip protectors would be deployed for clinical trials or deployed for use by vulnerable individuals who depend on the intervention.

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