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Introduction
Fall related arm injuries in children, such as those involving playground equipment, have a high incidence in New Zealand and other industrialised countries. Although there has been extensive biomechanical research into fall-related injuries in the elderly, we have found no reported investigations of this type concerning children. Computer simulation, using an impact model and a stochastic approach, may provide the means to estimate the impact and fracture forces associated with falls in children. Before a full investigation could be conducted, including the collection of data on children, a pilot study was required to test the feasibility of using the impact model.

Methods
A two-mass damper-spring computer model applied previously in jumping and fall research (van den Kroonenberg et al., 1995; Chiu et al., 1998; Nigg et al., 1999) was used to generate a temporal response curve of impact force at the wrist and shoulder of a single child during a fall impact on an outstretched hand. The model’s two masses, as shown in the centre of Figure 1 below, represents the mass of the arm segment (\(m_{\text{arm}}\)) that makes the initial contact with the ground and the mass of the torso and body segments (\(m_{\text{torso}}\)) suspended at initial contact. The effective stiffness and damping characteristics of wrist contact tissues (\(k_w\) & \(d_w\) respectively) and the shoulder (\(k_s\) & \(d_s\) respectively) are represented by spring and dashpot systems. The stiffness of the ground surface (\(k_g\)) is represented by a spring undergoing a deflection (\(x_g\)). The initial velocities of the arm and torso segments at impact (\(v_{\text{arm}}, v_{\text{torso}}\)) are dependent on the fall height (\(h\)). The impact forces were calculated at the wrist and shoulder with \(x_w\) and \(x_s\), representing the deflection of the joints. The equations of motion that govern the behaviour of the model are given in (Chiu et al., 1998).

To the direct right of the impact model in Figure 1 is a typical temporal curve of impact force at the wrist. The curve is characterised by a peak wrist force (\(F_1\)) of a high frequency transient at the moment of hand contact and a peak wrist force (\(F_2\)) of a low frequency transient as the torso deceleration force is transmitted through the arm. It is unknown at this time which force is most likely to cause fracture at the wrist. The impact force \(F_1\) is usually of higher magnitude than \(F_2\) (Chiu et al., 1998) but \(F_2\) occurs over a greater time and represents a greater energy transfer through the wrist. Therefore, both peak impact force values were used for the assessment of fracture risk in this study.

The frequency distribution curves on either side of the two-mass model in Figure 1 illustrate the stochastic input and output data for a given fall condition. A fall condition consisted of a child population group (defined by age range and fracture history), a fall height and an impact surface. The stochastic method involves multiple Monte Carlo simulations, each with a different random child or ‘case’ belonging to the population group. The results of a set number of cases for each fall condition were recorded and compared with other fall conditions.

The steps involved in stochastic simulation of fall impact are: 1) the generation of a ‘virtual’ case from a set of random values; 2) the simulation of the fall of this child for a given fall condition; and 3) the recording of peak impact forces that characterise this condition. The process was repeated for a predetermined number of cases (i.e. sample) for each fall condition, to generate an estimated statistical distribution of the ‘population’ peak impact forces. As shown on the left of Figure 1, each random value is sampled from a normal distribution with a mean value and standard deviation. The variables of interest included the segment masses, \(m_{\text{arm}}\) and \(m_{\text{torso}}\), and the joint properties \(k_w, d_w, k_s, d_s\). Each child population group have representative means and standard deviations of these variables. In the simulation,
the variables fall height, $h$, and ground stiffness, $k_g$, were set constant for each fall condition. Simulink® and MATLAB® software (MathWorks, Natick, MA, USA) were used to model the dynamics of the two-mass model and the stochastic simulation.

As shown on the right of Figure 1 the statistical distribution of the peak impact forces from multiple simulations are compared to the fracture force. The impact forces may not be normally distributed so the probability of fracture for a specific fall condition was calculated as the proportion of cases whose impact force exceeded the fracture force. By way of illustration, in Figure 1 about 70% of $F_1$ cases appear to exceed $F_{fracture}$ and 30% of $F_2$ exceed $F_{fracture}$. The effect of selected risk factors on the probability of fracture was tested by varying each factor separately from a baseline fall condition. In this study, the relative risks ($r_1, r_2$) of each factor were defined as the magnitude of change of the fracture proportions ($%F_1, %F_2$) from the baseline.

The compressive fracture force of the distal radius ($F_{fracture}$) was estimated as a function of apparent bone density, strain rate, and cross-sectional area (Carter et al., 1977; Gilbert et al., 1989; Carter et al., 1992). Population means and standard deviations for arm and torso masses and distal radius density and cross-sectional area were based on published data for female children (Goulding et al., 1998). The wrist and shoulder properties were based on adult data (Chiu et al., 1998) because of the lack of published data on children. The risk factors tested were fracture history, impact surface type, fall height, and age.

**Results & Discussion**

A control group aged 3-7 years (no fracture history) with a fall height of 1.5 m onto a 7 cm foam impact surface, was taken as the baseline condition. The strain rate was set at 0.2 s$^{-1}$ and 1000 simulations were generated for each condition. At baseline, 29% of controls had a $F_1 > F_{fracture}$ (i.e. $r_1\{29\%\}=1.0$ ) and 31% had a $F_2 > F_{fracture}$ (i.e. $r_2\{31\%\}=1.0$) (See Figure 2a below). Compared to baseline, a history of fracture increased the risks to $r_1\{83\%\}=2.86$ and $r_2\{78\%\}=2.52$ (See Figure 2b below); falling onto a rigid surface to $r_1= 3.41$ and $r_2=1.42$; increasing the fall height to 2.5 m to $r_1= 3.35$ and $r_2= 2.97$; and
increasing the child age to 8-10 and then 11-15 decreased the risks to $r_1 = .03$, $r_2 = .58$ and $r_1 = .00$, $r_2 = .02$ respectively.

![Figure 2a) Comparison of wrist impact temporal force to distal radius compressive strength and effect of fracture history: 2a) 3-7 years, no fracture history, 2b) 3-7 years, fracture history](image)

The effects of fracture history, surface type and fall height on risk of fracture are consistent with the injury prevention literature. Goulding et al. (2000) found that children with a previous fracture history were 3.28 times more likely to fracture their arms again over a 4 year period compared to non-fracture controls. (compared to $r_1 = 2.86$ and $r_2 = 2.52$ in this study). Robinovitch et al. (1998) reported that a thick foam pad reduced the first peak force by 2 to 3 times but had no effect on the second peak force ($r_1 = 3.41$ and $r_2 = 1.42$ in this study). Finally, Chalmers et al. (1996) reported that falls from heights over 1.5 meters were at 4.1 times risk of injury compared to falls below 1.5 meters ($r_1 = 3.35$ and $r_2 = 2.97$ for 2.5 m in this study). This study also found that the youngest age group was at dramatically greater risk than the older age groups. Younger children are likely to have lower stiffness and damping values than older children and therefore may be at lower risk than estimated. Further research, including the collection of child data, will help determine the real effect of age. This study confirmed the feasibility of the impact model and justifies a full investigation.

References

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