



Short communication

A single-spring model predicts the majority of variance in impact force during a fall onto the outstretched hand

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ABSTRACT

The objective of this study was to validate a single-spring model in predicting measured impact forces during an outstretched arm falling scenario. Using an integrated force plate, impact forces were assessed from 10 young adults (5 males; 5 females), falling from planted knees onto outstretched arms, from a random order of drop heights: 3, 5, 7, 10, 15, 20, and 25 cm. A single-spring model incorporating body mass, drop height plus the estimated linear stiffness of the upper extremity (hand, wrist and arm) was used to predict impact force on the hand. We used an analysis of variance linearity test to test the validity of using a linear stiffness coefficient in the model. We used linear regression to assess variance (R^2) in experimental impact force predicted by the single-spring model. We derived optimum linear stiffness coefficients for male, female and sex-combined. Our results indicated that the association between experimental and predicted impact forces was linear ($P < 0.05$). Explain variance in experimental impact force was $R^2 = 0.82$ for sex-combined, $R^2 = 0.88$ for males and $R^2 = 0.84$ for females. Optimum stiffness coefficients were 7436 N/m for sex-combined, 8989 N/m for males and 4527 N/m for females. In conclusion, a linear spring coefficient used in the single-spring model proved valid for predicting impact forces from fall heights up to 25 cm. Results also suggest the use of sex-specific spring coefficients when estimating impact force using the single-spring model. This model may improve impact force to bone strength ratios (factor-of-risk) and prediction of forearm and wrist fracture.

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1. Introduction

Forearm or wrist fractures occur when the force applied to bone exceeds bone strength (i.e., failure load). The most common scenario for this fracture-type is a fall onto the outstretched hand (FOOSH). A FOOSH represents the worst-case falling scenario since the force applied to the hand is high when impacting with a straightened arm and outstretched hand (Burkhart and Andrews, 2010; Chiu and Robinovitch, 1998). Previous research proposes that the factor-of-risk (ϕ) for a fracture can be determined using the ratio of applied force to bone failure load, with $\phi \geq 1$ indicating propensity for fracture (Hayes et al., 1991). Imaging combined with finite element modeling (MacNeil and Boyd, 2008; Pistoia et al., 2002) has enabled non-invasive estimates of bone failure load, while mathematical models are used to estimate the applied impact force from falling.

The main mathematical models used in estimating impact force include: single-damper (Chiu and Robinovitch, 1998); two-mass, spring-damper (Chiu and Robinovitch, 1998); and multi-component, spring-damper models (DeGoede and Ashton-Miller, 2003). Although inferior to the advanced multi-component spring-damper models, the single-damper model is most commonly applied for estimating impact forces (see Table 1 in (Kawalilak et al., 2014) for a detailed summary). This is likely due to the simplicity of the model. In equation form, the single-damper model is comprised as follows: $F_{\text{impact}} = C\sqrt{2gh}$, where F_{impact} is the predicted impact force, C is the damping coefficient (determined experimentally), g is acceleration due to gravity, and h is the falling height (generally height of the body's center of mass, typically estimated as $\frac{1}{2}$ standing height). In our previous research (Kawalilak et al., 2014), the single-damper model explained 61–81% of the variance in experimentally-derived impact forces from fall heights upwards of 25 cm.

A key limitation associated with the single-damper model is that it only accounts for participant height, and ignores participant mass. This limitation may explain moderate impact force

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predictions previously found with this model (Kawalilak et al., 2014). An alternative, simple model for predicting impact force is the single-spring model (Robinovitch et al., 1991), which accounts for both participant height and mass. This model estimates the impact force through the incorporation of a stiffness coefficient for the impacting structure. The single-spring model (including the associated stiffness coefficient) has been validated at the hip (Robinovitch et al., 1991), but the model has not yet been validated for the hand.

The objective of this research was to validate a single-spring model in predicting experimentally-derived impact forces during an outstretched hand falling scenario. Validation consisted of (1) determining whether a linear stiffness coefficient could be applied to estimate impact forces; (2) characterizing explained variance in impact force predicted by the model; and (3) identifying the linear stiffness coefficient which best-predicted impact force.

2. Methods

Study methodology was presented in detail in (Kawalilak et al., 2014), which focused on validating a single-damper model for predicting impact forces during a falling scenario. Study methodology is discussed briefly below. The novelty of this study pertains to the use of a single-spring model for predicting impact forces.

2.1. Participants

We recruited a convenient sample of 10 (5 males, 5 females) healthy adults (mean age: 26, SD 4 years) to participate in the study (height: 169.5, 9.5 cm; mass: 74.0, 14.3). The University of Saskatchewan Biomedical Research Ethics Board approved this study.

2.2. Fall-release apparatus

A fall-release apparatus suspended the participant's trunk into the air (Fig. 1). Participants were restrained in a harness and suspended above the ground via an electromagnet. Participant's dominant hand was aligned over the center of the force plate. To reduce participant anxiety associated with falling, the force plate was covered with a thin foam mat (1 cm thickness). This mat had only a slight effect on resulting impact forces due to minimal damping capacity (damping ratio = 0.08).

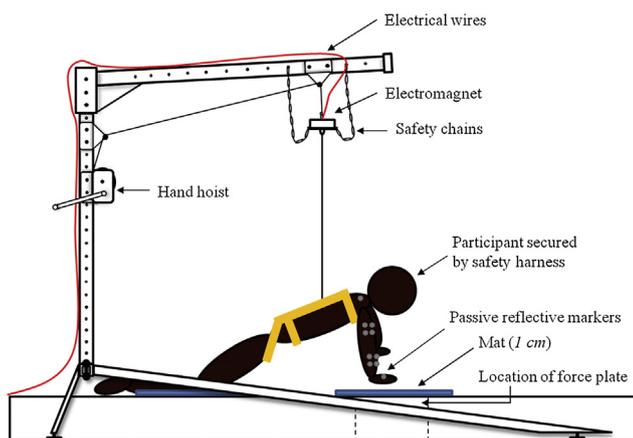


Fig. 1. Experimental set-up of the fall-release apparatus and falling scenario. The participant was positioned so that the outstretched hand was at the defined fall height directly over the force plate. The participant was supported using a fall-restraint harness connected to the fall-release apparatus via an electromagnet. The electromagnet was randomly timed to release the participant into a free-fall towards the force plate. From Kawalilak et al., 2014, with permission.

2.3. Determination of fall height

A motion capture system (Vicon Nexus, Vicon Motion Systems) was used to track the 3-dimensional coordinates of passive reflective markers located on the participant's body. Fall heights were determined by calculating the difference between the participant's hand and the top of the mat placed on top of the force plate. Motion data were captured at 200 Hz. We used the following nominal fall heights: 3, 5, 7, 10, 15, 20, and 25 cm (25 cm was the maximum capacity of our apparatus). Fall heights were randomized for each of the fall trials and for each participant. In total, each participant underwent 28 falls.

2.4. Body positioning

Participants were positioned on their knees, with their shins and dorsal/superior surface of their foot in contact with the ground, with their dominant hand and forearm outstretched in front of them over the force plate (Fig. 1). With this setup, ~50% body mass rested against the ground with ~50% body mass supported by the harness. This mass distribution was verified in the laboratory and matched prior research (Chiu and Robinovitch, 1998). Right and left hands were staggered, with the non-dominant arm slightly bent and raised relative to the outstretched dominant arm, as per (Chiu and Robinovitch, 1998). Rationale for staggering the hands was to ensure that the dominant hand impacted first. Participants were instructed to fully extend the elbow of their dominant arm during the fall and maintain this position during impact to simulate the worst-case falling scenario.

2.5. Experimental fall procedure

A random delay ranging from 0.1 to 5 s occurred before releasing the electromagnet suspending the participant.

2.6. Impact force assessment and force profile

The force at impact and static equilibrium was recorded using a force plate (OR6-7, AMTI). As per the Nyquist theorem, we sampled the force data at 2000 Hz since the amplifier applied a low-pass filter with a fixed cut-off frequency of 1000 Hz. Since power spectral analyses indicated no signals above 200 Hz, we did not apply any further filtering.

Experimental data resulted in a force profile with two peak forces (F_{1max} and F_{2max}) and a static equilibrium force (F_{static}) (Fig. 2). We used F_{1max} as the predicted impact force occurring on the dominant hand at impact. We used F_{static} to define the felled mass via $mass = 2F_{static}/gravity$. A factor of 2 was applied because, at equilibrium, each hand supported 50% of the felled mass resting against the force plate. Using custom software (Matlab 2006b, MathWorks), F_{1max} and F_{static} were defined for each fall trail. The average peak impact force for all fall trials, for each fall height, was calculated for each participant. The average F_{static} for all fall trials was calculated for each participant.

2.7. Mat adjustment

To account for the mat being placed upon the force plate (which lowered impact forces), we multiplied the mat's damping coefficient (75 Ns/m) by the velocity just prior to impact (as identified from the force plate data) and added this value to F_{1max} . This approach was verified using in-house impact studies with and without the mat.

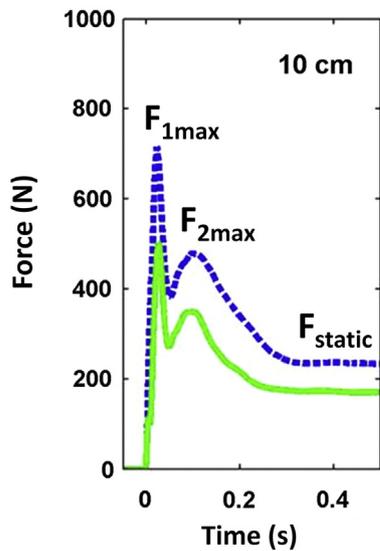


Fig. 2. An example of experimental force (N) versus time (s) profile illustrating the initial peak (F_{1max}) that occurred shortly after initial hand contact, followed by a second peak (F_{2max}) which is the force to decelerate the rest of the body mass. The static, stationary force (F_{static}) was used to define the felled mass (via $F_{static} = \text{mass} \times \text{gravity}$). The blue (dashed) line represents the average male impact force at this drop height (10 cm); green (solid) line represents the average female impact force at this drop height (10 cm). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

2.8. Single-spring model

Impact force (F_{impact} , N) was estimated using the following equation (Robinovitch et al., 1991):

$$F_{impact} = \sqrt{2hmgk}$$

where h was the drop height (m), m was mass (kg), g is the gravitational constant (9.81 m/s^2), and k is the stiffness constant for the upper extremity (hand, wrist and arm) (N/m). For this research, we set $F_{impact} = F_{1max}$ and $2F_{static} = mg$.

2.9. Analysis

To test the validity of using a linear stiffness coefficient in the model, we assessed the linearity between experimental and predicted impact forces using an analysis of variance linearity test. We used linear regression to assess variance (R^2) in experimental impact force predicted by the single-spring model. We used back-projection to estimate the linear stiffness coefficient k used in the single-spring model. Specifically, k was adjusted until the regression slope and intercept between experimental and predicted impact force equaled 1 and 0, respectively. Separate linear stiffness coefficients were derived for male, female and sex-combined. Significance was set to $\alpha = 0.05$. Statistical analyses were performed using commercial software (PASW, Version 22, SPSS Inc).

3. Results

The association between experimental and predicted impact forces was linear ($P < 0.05$); as such, usage of a linear stiffness coefficient in the single-spring model was valid. Explained variance was $R^2 = 0.82$ for the sex-combined model, $R^2 = 0.88$ for males and $R^2 = 0.84$ for females (Fig. 3). Optimum stiffness coefficients were $k = 7436 \text{ N/m}$ for sex-combined, $k = 8989 \text{ N/m}$ for males and $k = 4527 \text{ N/m}$ for females.

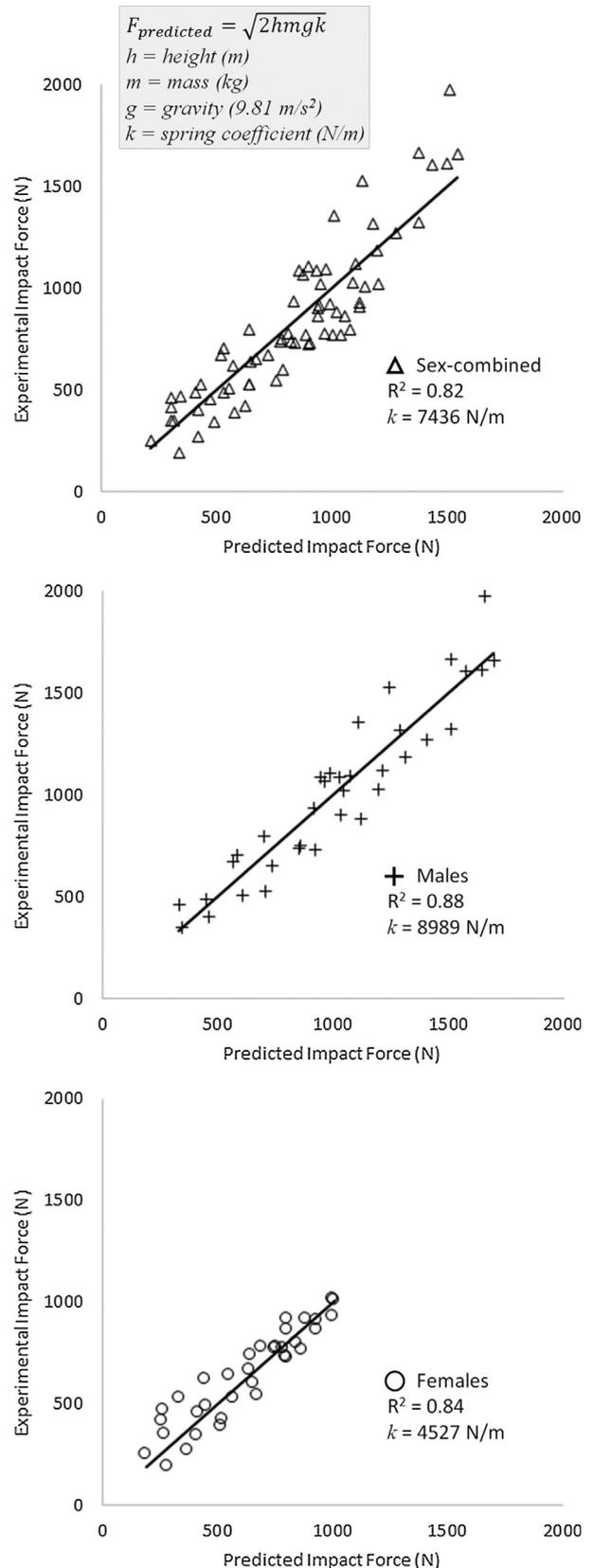


Fig. 3. Linear regression between experimentally measured impact force and single-spring-model predicted impact force. The triangle data points represent combined sexes; star data points represent males; and circle data points represent females. Reported k stiffness values coincide with the equation at the top of the figure. For estimating impact force from standing height, h is typically defined as the height of the body's center of mass, often estimated as half an individual's standing height.

4. Discussion

This study validated a single-spring model in predicting measured impact forces during an outstretched arm falling scenario. To our knowledge, this is the first study to validate a single-spring model of the arm.

We previously validated a single-damper model using the same data set (Kawalilak et al., 2014). This model explained 61–81% of the variance in experimentally-derived impact forces (sex combined: $R^2 = 0.61$; males: $R^2 = 0.70$; females: $R^2 = 0.81$). The single-spring model, conversely, better fitted experimental data, explaining more variance in experimental impact forces (sex-combined: +21%; males: +18%; females: +3%). The key difference with the single-spring model is that it accounted for both participant height and mass. Of note, more advanced models, such as the two-mass, spring-damper model proposed by (Chiu and Robinovitch, 1998), also take into account height and mass, as well as sex. Such models are capable of estimating $F_{1\max}$, $F_{2\max}$, and the force transferred to the shoulder. The single-spring model, conversely, is only capable of estimating $F_{1\max}$. A key benefit though of the single-spring model (like that of the single-damper model) is its simplicity and ease of application for estimating factor-of-risk for forearm and wrist fractures.

This study found that males and females had different model coefficients. This finding is supported by previous reports (Chiu and Robinovitch, 1998; Kawalilak et al., 2014). As such, when using the single-spring model, we recommended usage of sex-specific spring coefficients when estimating fracture risk. Importantly, the use of a general spring coefficient may over-estimate fracture risk, specifically for females. As females have higher risk for osteoporotic wrist fracture, over-estimate of fracture risk may lead to unnecessary treatment and detrimental side effects (e.g., gastrointestinal problems (Brown et al., 2014; Pazianas et al., 2010)).

This study has specific limitations warranting discussion. First, although we report fall data up to 25 cm, this maximum fall height is $\sim 1/4$ the height of a person's center of mass. Future research is needed to validate the derived single-spring model at fall heights up to half the participant's standing height. Second, we estimated the subject's mass impacting the force plate from the static, stationary portion of the force plate profile data. However, a portion of the impacting mass may have been supported at the knees, leading to mass under-estimation. This is problematic as under-estimation of mass would lead to over-estimation of the derived stiffness coefficients, which would lead to over-estimation of impact force. To address this issue, we performed additional testing with passive forward falls and found that $\sim 50\%$ bodyweight was supported at the knees (matching prior research (Chiu and Robinovitch, 1998)) while each hand supported $\sim 25\%$ bodyweight. As our experimental data indicated an average F_{static} of 27% bodyweight, we believe our mass estimations and spring-coefficients are representative. Third, our sample of males and females was small. It would be prudent to cross-validate the proposed models

with a larger sample. Fourth, our sample is comprised of young adult males and females. As such, derived spring-coefficients are specific to a young population.

In conclusion, our results indicate that the single-spring model proved valid for predicting impact forces from fall heights up to 25 cm. Our results also suggest the use of sex-specific spring coefficients. The single-spring model is a potential alternative to the single-damper or multi-component spring-damper models for predicting impact forces. We conclude that investigators concerned with determining the factor-of-risk for forearm and wrist fractures associated with a FOOSH should consider using single-spring models proposed here.

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Conflict of interest

There are no conflicts of interest to declare.

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