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Estimating severity of sideways fall using a generic multi linear regression model based on kinematic input variables

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ABSTRACT

Many research groups have studied fall impact mechanics to understand how fall severity can be reduced to prevent hip fractures. Yet, direct impact force measurements with force plates are restricted to a very limited repertoire of experimental falls. The purpose of this study was to develop a generic model for estimating hip impact forces (i.e. fall severity) in *in vivo* sideways falls without the use of force plates.

Twelve experienced judokas performed sideways Martial Arts (MA) and Block ('natural') falls on a force plate, both with and without a mat on top. Data were analyzed to determine the hip impact force and to derive 11 selected (subject-specific and kinematic) variables. Falls from kneeling height were used to perform a stepwise regression procedure to assess the effects of these input variables and build the model.

The final model includes four input variables, involving one subject-specific measure and three kinematic variables: maximum upper body deceleration, body mass, shoulder angle at the instant of 'maximum impact' and maximum hip deceleration. The results showed that estimated and measured hip impact forces were linearly related (explained variances ranging from 46 to 63%). Hip impact forces of MA falls onto the mat from a standing position (3650 ± 916 N) estimated by the final model were comparable with measured values (3698 ± 689 N), even though these data were not used for training the model. In conclusion, a generic linear regression model was developed that enables the assessment of fall severity through kinematic measures of sideways falls, without using force plates.

1. Introduction

Femoral fractures in the elderly are an important worldwide public health issue (Cheng et al., 2011). Most hip fractures are caused by falls; in particular by falls in the sideways direction with impact directly on the hip (Parkkari et al., 1999). Over the last few decades, several research groups have studied impact mechanics to understand how fall severity can be reduced.

Experimental and computer simulation studies have indicated that several fall strategies may substantially reduce fall severity (Groen et al., 2007; Lo and Ashton-Miller, 2008). However, evaluation of the protective effects of fall strategies in *in vivo* falls is challenging. A fundamental variable for fall severity is the load applied to the femoral bone during impact (van den Kroonenberg et al., 1995; Hayes et al., 1996). In experimental *in vivo* fall studies, fall impact load was defined by the peak impact forces measured by force platforms (Sabick et al., 1999; Nankaku et al., 2005; Groen et al., 2007; van der Zijden et al., 2012). Due to safety reasons, however, these studies are limited in the repertoire of experimental falls (e.g. low fall heights) for which impact forces can be measured directly.

Alternatively, various indirect measures have been used for estimating fall severity in, for instance, experimental falls from standing height on padded surfaces (Robinovitch et al., 2003). Based on an undamped single-degree-of-freedom mass-spring model, the hip impact velocity is considered to be a determinant for fall severity, i.e. hip impact force (van den Kroonenberg et al., 1995). Indeed, an experimental study (Groen et al., 2008) has shown moderate linear relations between hip impact velocity and impact force. However, these relations depended on fall technique. The impact location (Hsiao and Robinovitch, 1998; Smeesters et al., 2001; Robinovitch et al., 2003) and the trunk angle at impact (van den Kroonenberg et al., 1996; Groen et al., 2007)

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have also been used for evaluating fall severity. Trunk angle was proposed to reflect the effective mass of the falling body prior to impact, which associates with fall severity (van den Kroonenberg et al., 1995). In addition, loading configurations of the femoral bone (force direction and point of application) have been considered (van der Zijden et al., 2012).

Subject-specific measures like body mass and height have also been shown to affect hip fracture risk (Hayes et al., 1996). In addition, a higher body mass index (BMI) was reported to decrease the risk for hip fractures (Greenspan et al., 1994), which may be explained by an increased energy absorption by soft tissues in the pelvic region during impact in individuals with a high BMI (Bhan et al., 2014). Yet, the complex relationships between the various kinematic and subject-specific variables and the hip impact force remains incompletely understood, hindering the ability to estimate the latter from the former.

The purpose of this study was to develop a generic model including a limited number of kinematic and subject-specific variables for estimating hip impact forces (i.e. fall severity) in *in vivo* sideways falls without the use of force plates. The model was trained on a data set including two distinct fall strategies from *kneeling* position on a padded and on an unpadded surface. It was subsequently validated on a set of sideways falls from *standing* position, by comparing the estimated hip impact forces to measured force data of these falls. The rationale behind this approach is that if proven sufficiently accurate, such a model could be applied for determining hip impact severity in experimental falls from standing height onto thick safety mattresses that preclude the use of force plates.

2. Methods

2.1. Participants

Twelve participants were recruited from a local judoka club (mean ± SD, age: 27.6 ± 10.7 years, body mass: 77.9 ± 12.2 kg, height: 1.80 ± 0.07 m, men-women: 9–3). All were healthy and had at least 10 years of judo experience. Each participant signed an informed consent form prior to participation. The protocol was approved by the Ethical Board of the region Arnhem-Nijmegen.

2.2. Experiment

Kinematic and force data were obtained from *in vivo* fall experiments as described in more detail previously (van der Zijden et al., 2012). A total of 33 reflective markers were attached to anatomical landmarks on the upper body, thigh and pelvic segments; after static calibration 17 of these markers were removed. Kinematic data were recorded with an eight-camera 3D motion analysis system (Vicon®, Oxford, UK) at 200 Hz. Ground reaction forces were measured with a force plate (1200 × 1200 mm, Bertec® Corporation, Columbus, USA) at 2400 Hz. After a calibration series, three fall series were recorded in which the participants performed two distinct fall strategies: the Block (a natural fall arrest strategy) and the Martial Arts (MA) technique (Fig. 1) (Groen et al., 2007; Weerdesteyn et al., 2008). In the Block technique, the outstretched ipsilateral arm is used to block the impending fall. Using the MA technique, the fall is converted into a rolling movement to distribute the impact energy over a greater contact area. The rolling movement is facilitated by trunk lateral flexion and rotation and shoulder protraction. After impact, the arm is used to break the fall, by slapping the arm onto the landing surface. For all falls, the participants started from a position next to the force plate and then fell on the force plate with both the lower and upper body parts. After fall initiation, an auditory cue (one syllable word) instructed the participant which fall technique he/she had to perform. In fall series A (with mat; 1200 × 1200 × 40 mm polyurethane foam, Agglorex®, Lommel, Belgium) and B (without mat) the participant performed 10 Block and 10 MA falls from a kneeling position in randomized order. In fall series C, the participant fell six times from a standing position using the MA technique. For safety reasons, no Block falls from a standing position were performed. The force plate was covered by the judo mat during series C as well. The sequence of the fall series was randomized across participants.

2.3. Data analysis

Kinematic data were analyzed using Matlab r2013a (The MathWorks Inc., Natick, USA). The virtual position of the left knee joint center, hip joint center and greater trochanter (LGT) marker were calculated using three reference markers on the left thigh segment and data of the calibration series (van der Zijden et al., 2012). Trials were excluded when markers required for analysis were occluded for more than 20 consecutive frames, or for more than 5 frames around impact. For marker trajectories with smaller gaps, cubic spline interpolation was applied. For series A, B and C, 14 (6%), 26 (11%) and 5 (7%) trials were excluded, respectively, from the analyses. Marker velocities and accelerations were calculated by the first and second derivatives of the unfiltered position data. Subsequently, the position, velocity and acceleration data were filtered with zero-lag 4th order Butterworth lowpass filters (cut-off frequencies of 100, 50 and 100 Hz, respectively).

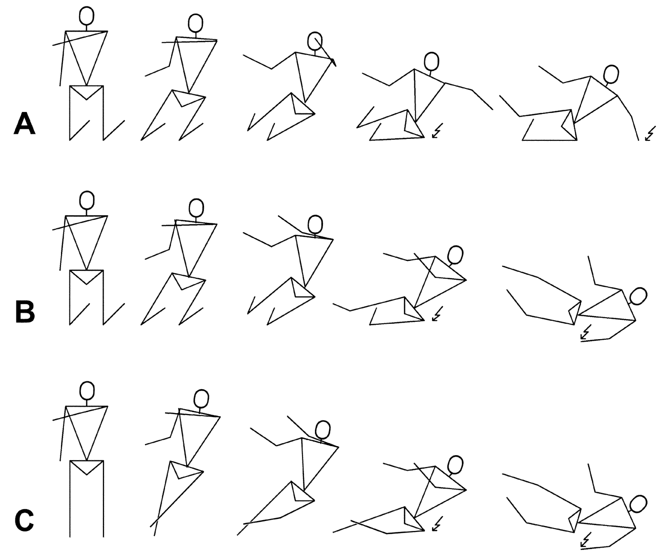

Fig. 1. Stick figures showing the falls from kneeling height and standing height (frontal view). Impact is indicated by a jagged arrow. (A) Block technique: the outstretched arm is used to block the impending fall. (B) Martial arts (MA) technique: the fall is converted into a rolling movement to distribute the impact energy over a greater contact area. (Reprinted from Groen et al. (2007), with permission from Elsevier.) (C) MA technique from standing height.

2.4. Estimating fall severity

The measured hip impact force was defined by the first peak of the vertical ground reaction force after the instant of impact force onset as measured by the force plate (Fig. 2D). Based on the kinematic data, two key impact events were identified to estimate the input variables. 'Start impact' was defined by the instant of peak downward velocity of the LGT marker (Fig. 2B) and 'maximum impact' by the instant of maximum LGT vertical deceleration (Fig. 2C), where deceleration corresponded to the slowing of downward motion.

A total of 11 subject-specific and kinematic variables were selected based on the literature and laws of physics. Kinematic variables were categorized as prior to, at and post-impact variables (see detailed definitions in Table 1 and Figs. 2 and 3).

- Subject-specific variables included body mass (BM) and body mass index (BMI).
- Prior to impact, we determined fall height (H) and peak hip impact velocity (HIPvel).
- At impact variables were hip impact deceleration (HIPdec) and peak time (Ptime), i.e. the elapsed time between the 'start impact' and the 'maximum impact' events.
- Furthermore, we included variables describing the impact posture, being the femur (FEMang_fr, FEMang_tr) and upper body (TRUNKang_fr, SHOUang) angles at 'maximum impact' in the frontal and the transversal plane.
- Post impact, we calculated the maximum vertical deceleration of the upper body (TRUNKdec).

2.5. Model design and validation

The input dataset for training the model involved all trials from series A and B (from kneeling position, with and without mat, $n = 440$). A stepwise linear regression model was built (Matlab r2015a, The MathWorks Inc., Natick, USA). Because possible non-linear relations were anticipated, we not only included linear terms, but also squared and two-way interaction terms of the 11 selected subject-specific and kinematic variables. The measured hip impact force (N) was set as the dependent variable (entry $p < 0.05$; removal $p > 0.10$).

The trained model was then validated on the data of series C (from standing position, with mat, MA only), i.e. data that had not been used to train/fit the model initially. The estimated hip impact force was calculated for each step in the model,