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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).

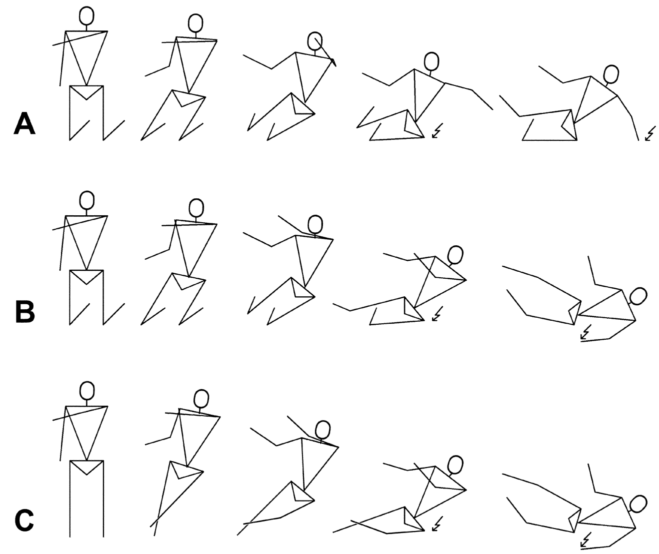

Fig. 1. Stick figures showing the falls from kneeling height and standing height (frontal view). Impact is indicated by a jagged arrow. (A) Block technique: the outstretched arm is used to block the impending fall. (B) Martial arts (MA) technique: the fall is converted into a rolling movement to distribute 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,

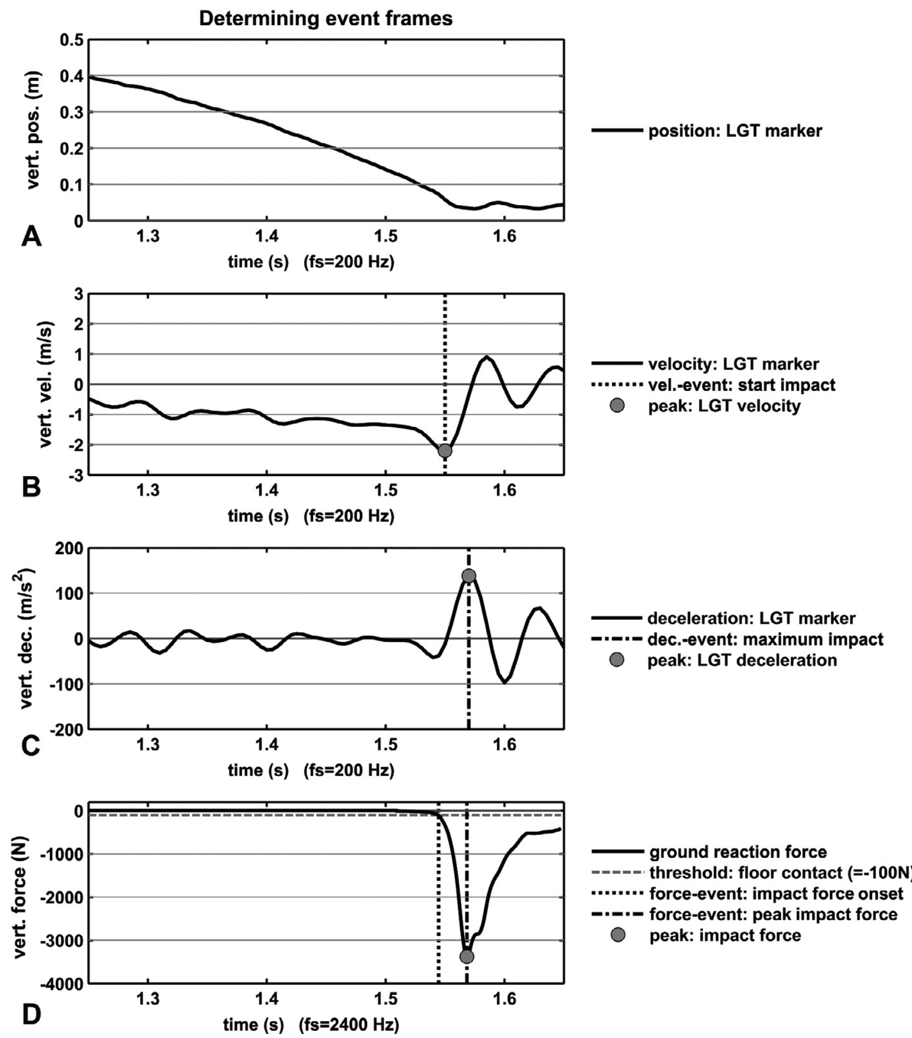

Fig. 2. Plots demonstrating determination of events by using kinematic data of the (virtual) marker on the left greater trochanter (LGT) (A–C) (sample frequency (fs) of 200 Hz) and force plate data (2D) (fs = 2400 Hz). First and second derivatives from the *z*-coordinate of the LGT marker (A) were calculated, with the *z*-axis being the vertical axis of the lab coordinate system pointing upwards. The 'start impact' and the 'maximum impact' events were estimated by the instant of the peak downward (negative) velocity of the LGT marker (B) and maximum LGT vertical deceleration of the downward motion (C), respectively. Based on the force plate data, the instant of impact force onset (floor contact) was defined by the frame in which the vertical force exceeded a threshold of −100 N and the instant of peak impact force was defined by the first peak of the vertical ground reaction force (2D).

increasing the number of input variables systematically. The estimated and measured hip impact forces were compared using the R^2 and root mean square (RMS) errors calculated for each step. To prevent overfitting of the model, the number of variables in the final model was determined by the optimum in the estimated R^2 and RMS error curves for falls from standing position (series C).

Finally, the validity of the final model was evaluated by comparing the mean estimated and measured hip impact forces calculated for each series and technique, i.e. on both the training (series A&B) and validation (series C) data.

3. Results

The stepwise regression procedure selected 16 variables for inclusion in the model to predict hip impact force during series A and B (falls from kneeling height; Table 2). Fig. 4 shows the results of this procedure demonstrating an increasing R^2-value and decreasing RMS error of the trained model when more input variables were included. With increasing number of input variables, the R^2-values increased from 0.4115 when only one input variable (TRUNKdec) was included to 0.7664 when 16 variables were included (Table 2); the RMS error decreased from 619 N to 394 N (Fig. 4B).

For validation, the trained model (input variables and their coefficients) was applied to the data of series C (falls from standing position). A peak R^2-value of 0.4833 was reached when input variables x_1 to x_5 (respectively, TRUNKdec, BM, SHOUang, HIPdec and FEMang_tr) were included in the model (Fig. 4A). Yet, the lowest RMS error (674 N) was found when input variables up to x_4 were included into the model (Fig. 4B). Based on these results, the cut-off number of input variables for the final model was chosen at x_4, as RMS errors were relatively low and R^2-values relatively high (RMS errors of 496 N and 674 N and R^2-values of 0.6259 and 0.4592 for the training and validation procedures, respectively; Fig. 4). The final multi linear model was thus:

$$\text{Hip impact force(N)} = -326 + 31 \cdot \text{TRUNKdec}(\text{m s}^{-2}) + 24 \cdot \text{BM(kg)} - 11 \cdot \text{SHOUang}(^\circ) + 6 \cdot \text{HIPdec}(\text{m s}^{-2})$$

The mean estimated and measured hip impact forces for each series and technique are shown in Fig. 5. The average differences between the estimated and measured hip impact forces ranged between 42 N and 229 N.

Table 1
Input variables for the stepwise procedure and means and standard deviations per series (A: knee - mat; B: knee - no mat; C: stand - mat) and fall technique (Block or martial arts (MA)).

Variable	Measure	ID	Mean ± SD	
Body mass	Body mass	BM	77.9 ± 12.2 kg	
Body mass index (BMI)	Body mass index	BMI	24.2 ± 3.5 kg m^{-2}	
Fall height	Initial vertical distance between left greater trochanter (LGT) and floor surface at the beginning of each trial	H	Series A&B: 0.49 ± 0.04 m	Series C: 0.95 ± 0.07 m
Peak hip impact velocity	Peak downward vertical velocity of the LGT marker. Negative values correspond to a downward motion (Fig. 2B)	HIPvel	Block: A: −1.6 ± 0.4 m s^{-1} B: −1.5 ± 0.3 m s^{-1}	MA: A: −1.4 ± 0.3 m s^{-1} B: −1.3 ± 0.2 m s^{-1} C: −2.3 ± 0.5 m s^{-1}
Hip impact deceleration	Maximum vertical deceleration of LGT marker. Positive values correspond to slowing of the downward motion (Fig. 2C)	HIPdec	Block: A: 60 ± 17 m s^{-2} B: 85 ± 22 m s^{-2}	MA: A: 48 ± 17 m s^{-2} B: 69 ± 20 m s^{-2} C: 165 ± 84 m s^{-2}
Peak time	Elapsed time between the 'start impact' and the 'maximum impact' events (Fig. 2B and C)	Ptime	Block: A: 48 ± 15 ms B: 29 ± 4 ms	MA: A: 49 ± 11 ms B: 33 ± 8 ms C: 32 ± 6 ms
Femur angles	Orientation of femoral segment in the frontal (*YZ*) plane at 'maximum impact': angle between the line from left hip (LHJC) to knee (LKJC) joint center and the horizontal. Positive values correspond to abduction (Fig. 3A)	FEMang_fr	Block: A: 0 ± 4° B: −1 ± 3°	MA: A: 1 ± 4° B: −1 ± 3° C: 6 ± 10°
	Orientation of femoral neck axis in the transversal (*XZ*) plane at 'maximum impact': angle between the line from (virtual) LGT to LHJC marker and the vertical. Positive values correspond to internal rotation (Fig. 3C)	FEMang_tr	Block: A: 9 ± 12° B: 7 ± 12°	MA: A: 5 ± 12° B: 5 ± 12° C: 3 ± 12°
Upper body angles	Orientation of the trunk segment in the frontal (*YZ*) plane at 'maximum impact': angle between the line from midpoint of the ASIS markers (LASI, RASI) to clavicle (CLAV) marker and the vertical. Positive values correspond to a trunk orientation to the left (Fig. 3A)	TRUNKang_fr	Block: A: 16 ± 20° B: 7 ± 14°	MA: A: 31 ± 10° B: 15 ± 9° C: 23 ± 11°
	Orientation of the shoulders in the transversal plane at 'maximum impact' in relation to the sideways fall direction: angle between the line connecting left and right shoulder (LSHO, RSHO) markers and the *y*-axis. The angle was measured within a plane (*X′Y′*) oriented perpendicular to the longitudinal axis of the trunk in the frontal (*YZ*) plane. Positive values correspond to a forward rotation of the left shoulder (Fig. 3B)	SHOUang	Block: A: 34 ± 18° B: 36 ± 17°	MA: A: 56 ± 13° B: 54 ± 16° C: 56 ± 13°
Upper body deceleration	Maximum vertical deceleration of CLAV marker post 'maximum impact'. Positive values correspond to slowing of the downward motion	TRUNKdec	Block: A: 42 ± 17 m s^{-2} B: 35 ± 12 m s^{-2}	MA: A: 27 ± 14 m s^{-2} B: 26 ± 11 m s^{-2} C: 59 ± 23 m s^{-2}

4. Discussion

The purpose of this study was to develop a generic multi-linear model using stepwise regression including a limited set of subject-specific and kinematic variables for estimating hip impact forces, i.e. fall severity, without the use of force plates; and to test its validity on a data set not included in training the model. The proposed model finally included four variables: TRUNKdec, BM, SHOUang and HIPdec. The results showed that estimated and measured hip impact forces were linearly related. In addition, hip impact forces of falls from a standing position as estimated by the final model were in line with measured values, even though these data were not used for training the model (which procedure solely included falls from kneeling height). The final multi-linear model is generic, straightforward and deemed sufficiently accurate as it includes a limited number of input variables which together yield estimates of hip impact forces in various sideways falls; the explained variances ranged from 46% to 63% for the validation and training data sets, respectively.

The relation between distinct input variables and hip impact force is complex and yet incompletely understood. Instead of considering a detailed full-body kinematic or musculoskeletal model including all complex relations, laws and formulas, we applied a more pragmatic stepwise regression procedure considering the linear, quadratic and interaction terms of 11 selected (subject-specific and kinematic) variables, which resulted in the proposed final model.

Of the four input variables that were included in the model, the maximum vertical deceleration of the upper body post impact (x_1 = TRUNKdec(m s^{-2}), b_1 = +31) had the strongest contribution to the explained variance in hip impact forces (in N). This variable appears to reflect one of the differences between the Block and MA technique. In Block falls, the upper body stops moving abruptly very shortly after hip impact, yielding a large peak in upper body deceleration. In contrast, in the MA technique, the fall is converted into a rolling movement, which enables a smooth deceleration of the upper body and results in a lower peak deceleration of the upper body after impact, as shown by the positive b_1-coefficient. We hypothesized that differences in TRUNKdec could reflect differences in effective mass. As such, the greater inclination (TRUNKang_fr) and more gradual deceleration of the trunk in the frontal plane in the MA technique may indicate a reduced effective mass that results in a reduced hip impact force (van den Kroonenberg et al., 1995). It must be noted, however, that

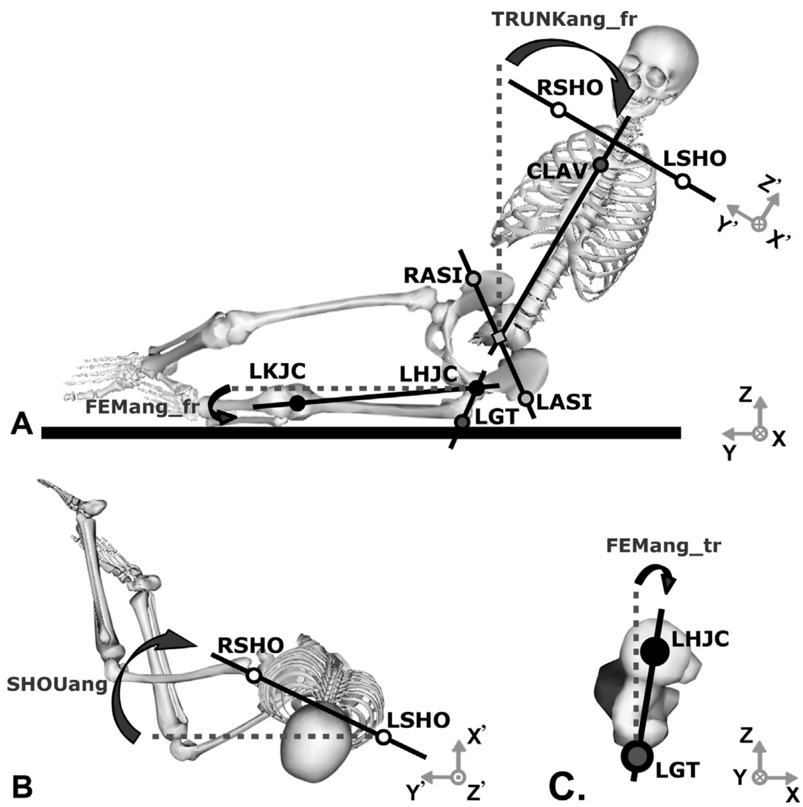

Fig. 3. Determining body configurations during the event of 'maximum impact'. (A) The frontal (*YZ*) plane: orientation of the femoral segment (FEMang_fr) was defined by the angle between the line connecting left knee (LKJC) and left hip (LHJC) joint centers (black dots) and the horizontal (dashed line). The lateral flexion angle of the upper body segment (TRUNKang) was defined as the angle between the line connecting the mean (light gray square) of the left (LASI) and right (RASI) anterior superior iliac spine markers (light gray dots) and the clavicle (CLAV) marker (dark gray dot) and the vertical (dashed line). (B) Rotation of the upper body (SHOUang) was defined as the angle between the line connecting left (LSHO) and right (RSHO) shoulder markers (white dots) and the fall direction (*y*-axis) in the $X'Y'$ plane (transversal plane corrected with the lateral inclination angle of the trunk). (C) Orientation of the femoral segment in the transversal plane (FEMang_tr) was defined by the angle between the line connecting left greater trochanter (LGT) (dark gray dot) and LHJC (black dot) and the vertical (dashed line).

Table 2
Output for the multi linear model using stepwise regression: ranking of the input variables (containing linear, squared and two-way interaction terms) based on the R^2-criterion for added value. The bold rows of the table represent the four input variables of the final model.

Coefficient ID	Input variable		Term type	R^2
$\boldsymbol{b_0}$		**Constant term**	**Constant**	
$\boldsymbol{b_1}$	$\boldsymbol{x_1}$	**TRUNKdec**	**Single**	**0.4115**
$\boldsymbol{b_2}$	$\boldsymbol{x_2}$	**BM**	**Single**	**0.5205**
$\boldsymbol{b_3}$	$\boldsymbol{x_3}$	**SHOUang**	**Single**	**0.5869**
$\boldsymbol{b_4}$	$\boldsymbol{x_4}$	**HIPdec**	**Single**	**0.6259**
b_5	x_5	FEMang_tr	Single	0.6615
b_6	x_6	BM * FEMang_tr	Interaction	0.6848
b_7	x_7	FEMang_tr * TRUNKdec	Interaction	0.7071
b_8	x_8	SHOUang * TRUNKdec	Interaction	0.7232
b_9	x_9	TRUNKang_fr	Single	0.7330
b_{10}	x_{10}	FEMang_tr^2	Squared	0.7406
b_{11}	x_{11}	HIPdec * FEMang_tr	Interaction	0.7477
b_{12}	x_{12}	HIPdec * TRUNKdec	Interaction	0.7524
b_{13}	x_{13}	Ptime	Single	0.7568
b_{14}	x_{14}	TRUNKang_fr * TRUNKdec	Interaction	0.7607
b_{15}	x_{15}	BM * Ptime	Interaction	0.7638
b_{16}	x_{16}	BM * SHOUang	Interaction	0.7664

Note: All resulting 16 input variables are shown in the table (*p*-values for variable entry and removal were $p < 0.05$ and $p > 0.10$, respectively). For definition of variable names see Table 1.

TRUNKdec occurs after hip impact. Hence, the association between TRUNKdec and hip impact force is likely reflective of true causal factors before or during hip impact, which we apparently failed to identify in our set of kinematic variables, instead of bearing a causal relationship in itself.

The second variable, body mass (x_2 = BM(kg), b_2 = +24) was expected to be important for the estimation of hip impact forces, as most laws of physics regarding potential, kinetic and impact energies involve the mass of a moving object. As such, body mass constitutes both a subject-specific measure and a measure reflecting the kinetic energy prior to impact.

The third variable that contributed importantly to the explained variance of the model was the shoulder orientation in the transversal plane (x_3 = SHOUang(°), $b_3 = -11$); it constitutes a composite

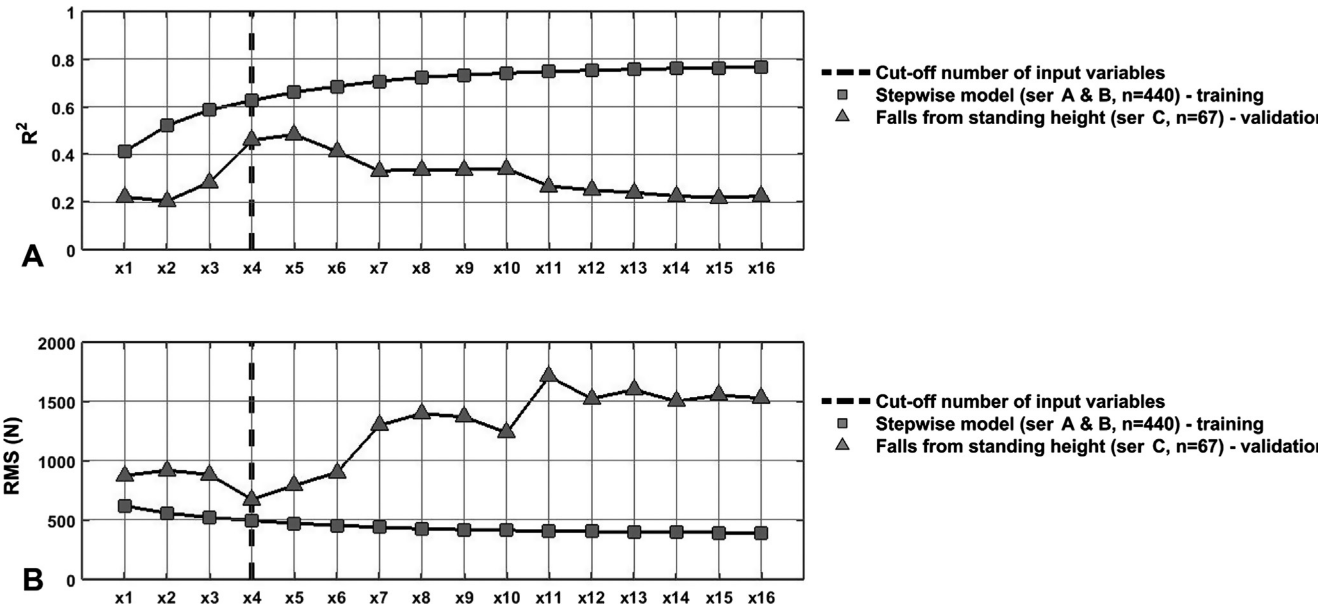

Fig. 4. Results of the multi linear regression model using stepwise regression (squares). The model was applied to the kinematic data of the martial arts (MA) falls from standing height, calculating the estimated impact forces in these falls (triangles) for each step of the procedure. R^2 (A) and RMS (B) values are shown for each step (i.e. increasing the number of input variables) comparing the estimated hip impact forces with the measured hip impact forces. For the final model, the cut-off number of input variables was chosen at 4× (black dashed line).

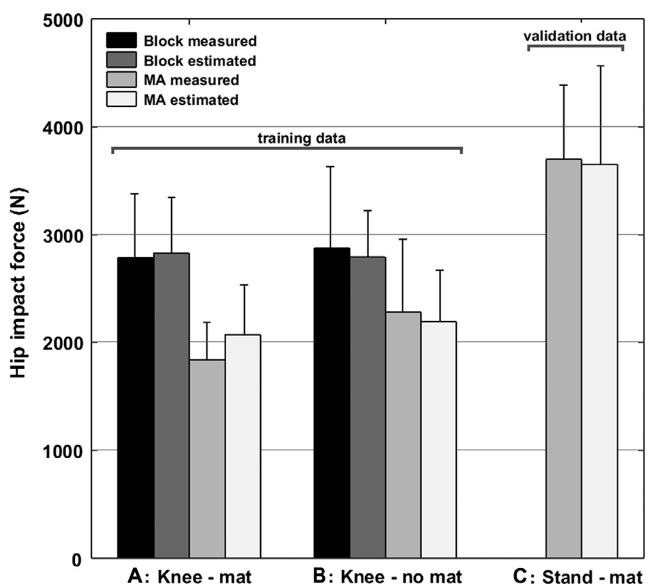

Fig. 5. The mean measured and estimated magnitudes of the impact force for all Block and martial arts (MA) falls within each series (from a kneeling position with (A) and without use of a mat (B) and from a standing position with use of a mat (C)). Error bars represents the SD.

measure of pelvis and trunk rotation (about the longitudinal body axis) at the instant of 'maximum impact'. Greater angles indicate more forward rotation of the left shoulder and associated with lower hip impact forces, which is shown by the negative sign of the b_3-coefficient. Large forward rotations of the left shoulder were particularly evident in MA falls (see Table 1), which observation is in line with previous findings of greater muscle activity in the pectoralis major in MA than in Block falls (van Swigchem et al., 2009). It may be suggested that SHOUang affects the hip impact forces by influencing the impact location in the hip area. However, our data were not in line with this hypothesis, as the major differences in SHOUang between MA and Block falls were not accompanied by similar differences in the FEMang_tr. Alternatively, we suggest that a greater rotational movement of the trunk along its longitudinal axis prior to impact (i.e. larger SHOUang) may have served to absorb some of the fall energy, leaving less fall energy to be absorbed at hip impact.

The fourth determinant in the model was the maximum vertical deceleration of the hip (x_4 = HIPdec(m s^{-2}), b_4 = +6). Greater deceleration associated with higher hip impact forces. Interestingly, HIPdec showed a stronger association than HIPvel. Hence, the direct linear relationship between HIPdec and hip impact force (as defined by Newton's second law F = ma), seemed to be stronger than the potential linear relation between HIPvel and hip impact force, as assumed in a perfect undamped single-degree-of-freedom mass-spring system impact model (van den Kroonenberg et al., 1995). Yet, it appears that the variability in HIPdec values (and consequently hip impact forces) not only originated from the differential execution of the requested fall techniques across trials and participants, but was also accounted for by falling on a padded as well as an unpadded force plate (series A and B, respectively). As can be seen in Table 1, HIPdec values in series A were considerably smaller than those in series B. The different fall surfaces that we used may therefore explain why this variable was included in the final model rather than HIPvel.

The current model yielded acceptable accuracy in estimating hip impact forces at both individual and group level reflected by RMS errors of 496 N and 674 N, respectively and average differences between the estimated and measured hip impact forces between 42 N and 229 N. Nevertheless, a substantial amount of variance remained unaccounted for. We suggest that further improvements in the accuracy of our model may be achieved by including subject-specific variables related to, for instance, anatomy and soft tissue thickness. The precision of the event detection based on kinematics may also be a source of variance. Differences between events derived from kinematic and from force plate data (gold standard) were 8.0 ± 22.5 ms for 'start impact' (versus impact force onset) and 8.5 ± 11.0 ms for 'maximum impact' (versus instant of peak impact force). Hence, on average the events were detected one or two frames later. In particular, the relatively large standard deviation for detection of 'start impact' points at substantial random errors in Ptime values (i.e. time between 'start impact' and 'maximum impact'), which may explain why this input variable did not significantly add to our model. Further sources of unexplained variance include the precision of the motion analysis system (<1 mm), and filtering and differentiation of the data.

Another limitation of our study pertains to the choice to validate the model using series C (MA falls from standing height), which did not include any data that corresponded to use of the block technique nor any data that corresponded to landing on an unpadded surface. This precludes drawing firm conclusions regarding the generalizability of our final model to fall scenarios other than those studied. Yet, we feel that our validation method provided the best possible evidence for the utility of the final model for estimating hip impact forces in falls from standing height, a position from which real-life falls usually happen. Nonetheless, in an additional analysis we verified the robustness of the model, by training it on a subset of 80% randomly selected trials from series A, B and C, with subsequent validation on the remaining 20% of trials. The trained model explained 64% of the total variance in hip impact forces (c.f. 63% for the proposed final model) and rendered the same top-four variables. The validation procedure yielded 69% explained variance (c.f. 46% for the proposed model). Hence, the results from this additional analysis confirms the robustness of the proposed model.

Furthermore, it is also unknown to which extent our model based on data obtained from experienced judoka's who performed experimental falls in a lab setting is applicable to other populations and conditions; its generalizability to fall-prone populations (e.g. elderly people) and more realistic falls scenarios (e.g. fall resulting from an unexpected slip during walking) remains to be investigated.

The potential applicability of the final model is twofold. Firstly, the model may be applied to estimate fall severity in experimental falls that do not allow the use of force plates, and may thus extend the repertoire of experimental sideways falls for which hip impact forces can be estimated with reasonable accuracy. This may involve assessment of falls from standing height that require thick padding of the floor surface for safety reasons. The model is expected to be useful for evaluating the effects of various fall strategies and of fall training interventions on fall severity in experimental settings. Yet, it must be noted that the inclusion of variables that pertain to events during and after impact imply that the fall of interest must actually occur, which is a limitation that precludes studying more realistic fall scenarios in older people. Secondly, rough estimates of the four variables in the final model can potentially be derived from (2D) video data, allowing application of the model in real-world falls. Recent studies have reported video footage of falls in older adults in long-term care (Robinovitch et al., 2013) and have yielded the first data on hip impact velocities from a small subset of these real-life fall recordings (Choi et al., 2015). Although the temporal and spatial resolution of this type of data may not be sufficient to provide exact outcomes, application of our model may provide relevant insights into (protective) fall strategies as applied by older adults in daily life.

In conclusion, we developed a generic multi-linear model using stepwise regression that enables fall researchers to assess fall severity through kinematic measures of sideways falls without using force plates. This study provides insights on how kinematic variables contribute to the estimation of hip impact forces in sideways falls. Future research employing this model may focus on examining impact mechanics of experimental sideways fall strategies onto thick safety mattresses captured with 3D kinematic data, or may focus on estimating impact severity of video-captured real-life falls.

Conflict of interest

The authors declare that they have no conflict of interest.

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