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Sideways fall-induced impact force and its effect on hip fracture risk: a review

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Abstract

Summary Osteoporotic hip fracture, mostly induced in falls among the elderly, is a major health burden over the world. The impact force applied to the hip is an important factor in determining the risk of hip fracture. However, biomechanical researches have yielded conflicting conclusions about whether the fall-induced impact force can be accurately predicted by the available models. It also has been debated whether or not the effect of impact force has been considered appropriately in hip fracture risk assessment tools. This study aimed to provide a state-of-the-art review of the available methods for predicting the impact force, investigate their strengths/limitations, and suggest further improvements in modeling of human body falling.

Methods We divided the effective parameters on impact force to two categories: (1) the parameters that can be determined subject-specifically and (2) the parameters that may significantly vary from fall to fall for an individual and cannot be considered subject-specifically.

Results The parameters in the first category can be investigated in human body fall experiments. Video capture of real-life falls was reported as a valuable method to investigate the parameters in the second category that significantly affect the impact force and cannot be determined in human body fall experiments.

Conclusions The analysis of the gathered data revealed that there is a need to develop modified biomechanical models for more accurate prediction of the impact force and appropriately adopt them in hip fracture risk assessment tools in order to achieve a better precision in identifying high-risk patients.

Introduction

Most injuries (87% of fractures) to older adults result from falls [1, 2]. Hip fracture is one of the most common consequences of falls [3, 4], as the applied force to the femur can easily exceed the bone strength. Hip fracture has become a major health problem over the world [5] as it is associated with significant morbidity, disability, reduced quality of life, and mortality [6]. It is also a substantial source of long-term institutionalization [7] and health care expenditure [8]. Statistical studies showed that falls from sideways are the main cause of hip fractures (63–69% in fall-related fractures) [5, 9]. Therefore, prediction of sideways fall-induced impact force to the femur has received much attention in the last two decades. There are two main approaches to predict the impact force for hip fracture risk assessment purposes: (a) mathematical modeling of human body falling and (b) empirical functions. In mathematical modeling, human body dynamic models are proposed to predict the kinematics of body falling, determine the impact velocity, and simulate the interaction between the body and the ground during the impact. Experimentally derived empirical functions are also proposed to predict the impact force based on the body parameters,

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mostly body weight, body height, and trochanteric soft tissue thickness (STT). Although remarkable efforts have been conducted to propose models for predicting the impact force, their accuracy is still questionable. The limited accuracy is due to inadequate consideration of joint reactions during the fall, ignoring the pre-impact movements of body segments, insufficient consideration of the effect of subject-specific fall type, fall direction (forward, backward, and sideways), and contact site on the magnitude of the impact force, and neglecting how the cause of fall (slip, trip, or uneven floor) and the initial conditions (fall from standing height, low stools, stairs, etc.) affect the kinematics of body segments during the fall. A considerable number of simplifications and assumptions have also been used in modeling, including limited number of links for analysis of the human body falling and insufficient number of degrees of freedom for simulating the impact phase by vibrational models.

Not only the available methods have limitations in predicting the impact force, impact force itself is not properly considered as a risk factor in hip fracture risk assessment tools. For example, the effect of impact force is completely ignored in areal bone mineral density (BMD) estimation from dual X-ray absorptiometry (DXA), volumetric BMD from quantitative computed tomography (QCT), and hip structural analysis (HSA). FRAX® (fracture risk assessment tool) considers the body weight and body height as rough representatives of the impact force. Nevertheless, the impact force is not explicitly calculated in FRAX. Only in 2D and 3D finite element analyses (FEA) of femur, the impact force is directly taken into account for assessment of hip fracture risk, although with a limited accuracy. This inadequate consideration may be due to the limitations of the available models for predicting impact force and insufficient understanding of the effect of real-life fall-induced impact force on hip fracture risk. Therefore, in spite of the importance of the impact force, it seems it has not received enough attention. This study aims to provide a review on researches invested on predicting the fall-induced impact force to (1) reveal major challenges and limitations in modeling of human body falling and characterization of the interaction between the body and the ground during the impact, (2) shed lights to resolve the current issues and clarify the missing parameters in the available models, and (3) provide suggestions for future research studies. To this end, the section "Effect of impact force on hip fracture risk" investigates the range of the fall-induced impact force and the range of the force that can cause a hip fracture. It also presents the potential reasons why impact force has not been appropriately considered as a risk factor in hip fracture risk assessment tools. The section "How much impact force differs from subject to subject" surveys the parameters that affect the impact force and evaluates the necessity for subject-specific prediction of impact force. The section "Available models to predict the impact force" provides a review on available models for

predicting the impact force and addresses their strengths/limitations. The section "Missing/neglected variables in available models for predicting fall-induced impact force" investigates the missing parameters in the available models and provides recommendations for future research studies.

Effect of impact force on hip fracture risk

From the biomechanical point of view, the fracture risk depends on the bone strength and the applied force. The fracture risk is measured by load to strength ratio (LSR) [10, 11] as the following:

$$LSR = \frac{The \ applied \ load}{The \ bone \ strength} \qquad (1)$$

where the bone strength is the maximum force that the bone can withstand without fracture and the applied load is the fall-induced impact force to the hip. If the applied load is greater than the bone strength (LSR>1), the bone will fracture. In this criterion, the applied impact force to the femur is one of the two main determinants of fracture risk [12], and thus, many researches have been conducted to determine the range of the fall-induced impact force as well as the range of the force that can cause a hip fracture. These two parameters are investigated in the following sections.

Range of the fall-induced impact force to the hip

Fall experiments and dynamic models are the two main approaches that have been used to determine the range of the fall-induced impact force. Experimental studies have implemented three methods: (1) voluntary natural fall from standing height by young adults, (2) pelvis-release experiments to measure the damping properties of hip soft tissues and predict the impact force, and (3) fall from a kneeling position to measure the impact force in a low-severity fall. Apart from the experiments, dynamic models have been developed to determine the fall-induced impact force. First, the kinematics of body segments during the fall are simulated by the dynamic models to determine the hip impact velocity. The velocity is then used in an impact model to simulate the interaction between the body and the ground and determine the applied force to the hip. Table 1 provides a review on the studies conducted to determine the range of hip impact velocity and force in a fall.

Based on the benchmark works, conducted by Robinovitch and his colleagues, the average pelvis impact velocity in an unexpected sideways fall from standing height is 3.0 m/s with a standard deviation of 1.0 m/s [23], although it can be higher for some persons, even up to approximately 5 m/s [16]. There is a positive correlation between the impact velocity and impact force during lateral pelvic impact [13]. The peak impact

Table 1 Results from studies reporting the range of impact velocity and force in falls on the hip

Study	Mean (SD or range) age in years, sample size			Method	Range of femoral force (N)	Range of impact velocity (m/s)
	Men	Women	Mixed			
Robinovitch et al. [13]	28 (5), $n=7$	26 (6), $n=7$	27 (6), $n=14$	Dynamic and impact model	Fall height 0.7 m: 5600–8600	1.40–6.07
Kroonenberg et al. [14]	—	—	—	Dynamic and impact model/ experiments	3720–9990 dynamic model 2900–4260 experiments	3.35–4.34 dynamic model 2.47–2.93 Experiments
Robinovitch et al. [15]	$n=3$	$n=6$	77 (10), $n=9$	Impact pendulum experiments	4050–6420	—
Kroonenberg et al. [16]	23.7 (3.7), $n=6$	—	—	Natural sideways fall from standing height	—	2.14–4.79
Hayes et al. [17]	$n=6$	$n=6$	20–35, $n=12$	Fall experiments from a height of 0.7 m	Men: 6100–12,100 Women: 5050–6370	—
Robinovitch et al. [18]	$n=5$	$n=5$	25 (4), $n=10$	Dynamic models and pelvis-release experiments	1145–5288	—
Robinovitch et al. [19]	—	—	—	Impact pendulum experiments	1700–5600	1.16–2.58
Sanndler and Robinovitch [20]	—	—	—	Pendulum dynamic model	—	1.34–4.14
Robinovitch et al. [21]	—	23 (5), $n=22$	—	Sideways fall experiments	160–387 J backward rotation 6–291 J forward rotation	0.58–3.71
Robinovitch et al. [22]	—	24 (5), $n=23$	—	Fall experiments	—	3.3 ± 0.3
Feldman and Robinovitch [23]	$n=13$	$n=31$	21 (2), $n=44$	Sideways fall experiments	—	3.01 (0.83)
Laing and Robinovitch [24]	—	23.1 (2.4), $n=14$	—	Pelvis-release experiments	1004–3434	—
Levine et al. [25]	$n=14$	$n=14$	22.4 (2.7), $n=28$	Pelvis-release experiments	1415 ± 235	—
Choi et al. [26]	—	—	—	Pendulum experiments for simulating moderate falls	3,465 (143)	2.0
Nasiri and Luo [27]	39.8 (22.1), $n=50$	53.7 (19.9), $n=80$	48.3 (21.8), $n=130$	Subject-specific dynamic and impact model	1883–5317	3.39–4.31
Average across the whole studies	37.0 (10.3)	40.1 (7.7)	39.1 (8.1)	—	5200	3.0

force in a fall from standing height is ranged from 4050 to 6420 N [15]. It may be increased up to 8600 N for an average individual in a fall from a pelvis height of 0.7 m [13]. However, in a low-severity fall, the impact velocity is approximately 1.0 m/s and the impact force can vary from 1 to 2.5 kN [24]. Generally, impact velocity of 3.0 m/s (SD = 1) [23] and femoral force of 5200 N [13] are the mean values in an unexpected fall from sideways for an average individual.

There are many uncertainties in an unexpected real fall that can significantly affect the magnitude of the impact force, including fall direction (forward, backward, or sideways), initial conditions of fall, impact site (anterior/posterior side of the femur or greater trochanter), body stiffness, stiffness of the floor (hard wood or carpeted), and pre-impact movements to arrest the fall [21–23, 28]. Therefore, the range of the impact velocity and impact force may vary from subject to subject and even for one subject from fall to fall. For example, any attempt to recover balance, which is common among young adults, may remarkably affect the impact force. Feldman et al. [23] showed that even unsuccessful attempts to recover balance by stepping may affect the magnitude of the hip impact force. Even holding an object [29] may affect the fall kinematics and reduce the applied force to the hip. Since people may experience falls in different environments, they may react differently and thus the impact force will be different. Therefore, if not impossible, it is very difficult to integrate all aspects of an unexpected real fall in modeling or even experiments. Hence, the listed results in Table 1 provide a fair estimation of the force that a subject may experience in a fall from standing height.

Range of the force that can cause a fracture in a femur

Apart from the studies conducted to find the range of the fall-induced impact force, tremendous efforts have been devoted to determine the strength of the femur and evaluate whether or not the fall-induced impact force is sufficient to cause a hip fracture. Experimental tests and finite element analysis are the most common approaches in determining the femoral strength. Table 2 provides a review on the studies conducted to determine the femur strength and the range of the force that can cause a fracture in a femur.

The reviewed studies generally investigated the femoral strength in single-limb stance and fall loading configurations. It has been illustrated that the strength of femur is generally lower under fall loading conditions. From the biomechanical point of view, the potential reason is that a higher moment is applied to the femoral neck in the fall loading configuration, and thus, the fracture occurs under a lower force [37]. Across the listed studies that investigated the single-limb stance loading configuration, the median femoral strength was 7214 N. The average femoral strength under fall loading configuration was 3462 N with a standard deviation of 1520 N (ranged from

573 to 15,034 N) [66]. The structural capacity of the proximal femur is even sensitive to a slight variation in lateral versus posterolateral loading angle in a sideways fall, i.e., when the subject rolls forward/backward. Slightly rolling forward can increase the load required to fracture the femur [40]. Not only the loading configuration affects the femoral fracture force but sex is also an effective parameter on the bone strength. The femoral strength was reported to be lower in women than in men. For the studies that male and female data were separated, the strength of femur was 4027 ± 1464 N for men and 2771 ± 1136 N for women. Age is also an effective parameter in determining the strength of femur due to its direct correlation with BMD. For studies that reported age-specific values, the femoral strength was approximately 40% lower in older adults (2888 N for subjects with the mean age of 80.9 years vs. 4766 N for subjects with the mean age of 63.9 years). Loading rate is another effective parameter on the femur strength. Although there is still an ongoing discussion on how the femur strength is affected by the load rate [62], fracture force is generally reported to be greater at a higher load rate [38, 67, 68].

Figure 1 provides a summary of Tables 1 and 2 and compares the range of fall-induced impact force to the hip with the average femur strength. Concluded from Fig. 1, the median value of the femoral force in an unexpected fall from sideways for an average individual (5200 N) is well within or even beyond the range of the fracture force measured in experimental studies (3500 N). Therefore, an unexpected lateral fall to the hip has the potential of fracturing the femur, especially in the elderly. Accurate prediction of fall-induced impact force is thus of great significance in developing hip fracture risk assessment tools. However, most of the currently available tools in clinics ignore the effect of impact force or do not appropriately consider it. The following section investigates how the impact force is considered in clinical hip fracture risk assessment tools.

Consideration of/neglecting the impact force in hip fracture risk assessment tools

Among the four dominant biomechanical variables affecting fracture risk, i.e., risk of fall, impact force, bone quality, and bone geometry [12], the first two variables are the difficult ones to be accommodated in assessment of fracture risk.

There are four fracture risk assessment tools currently used in clinical studies, i.e., BMD (T-score) determined from DXA, hip structural analysis (HSA), FRAX (fracture risk assessment tool), and finite element analysis. Bone densitometry based on hip DXA is now a well-established method in clinical centers for monitoring osteoporosis. It is referred to as the "gold standard" by WHO. However, bone quality is the only effective parameter in assessing hip fracture risk by T-score. It means if the bone quality is the same for two subjects who may

Table 2 Results from studies reporting the range of femoral fracture force

Study	Mean (SD or range) age in years, sample size			Method (objective of the study)	Mean (SD)/range of fracture force (N)
	Men	Women	Mixed		
Smith [30]		50, n = 1		SF/OLS (determining the fracture force)	4314–18826[a] for OLS / 2668–15034[a] for SF
Phillips et al. [31]			21-89[a], n = 39	OLS (determine the fracture force)	2668–13344[a]
Dalen et al. [32]	n = 7	n = 54	67 – 80[a], n = 61	OLS (evaluate the effect of BMD on the femur strength)	1000–9750[a]
Leichter et al. [33, 34]	32 – 83[a], n = 9	56 – 80[a], n = 9	57 – 87[a], n = 36	OLS (determine the fracture force)	1597–12740[a]
Alho et al. [35]	n = 18	n = 18		OLS (determine the fracture force)	2750–9610[a]
Esses et al. [36]	72 (5.5), n = 4	73.7 (13.9), n = 4	72.9 (9.8), n = 8	OLS (determine the femur strength)	2000–9550[a]
Lotz et al. [37]	70.9 (9.7), n = 7	67.4 (7.7), n = 5	69.4 (8.7), n = 12	SF (determine the fracture force)	778–4039[a]
Courtney et al. [38]	n = 6[c] / n = 7[d]	n = 4[c] / n = 3[d]	73.8 (7.1)[c] / 31.7 (12)[d]	SF (evaluate the effect of load rate on the femur strength)	3450 (1340) static loading (2 mm/s) / 4170 (1590) dynamic loading (100 mm/s)
Bouxsein et al. [39]	n = 10	n = 6	76 (59–96)[b], n = 16	SF (identify the femoral fracture load)	3680 (1540)
Pinilla et al. [40]			78 (9.7), n = 33	SF (evaluate the effect of loading direction on the femur strength)	4050 (900) for 0° load angle / 3820 (910) for 15° load angle / 3060 (890) for 30° load angle
Cheng et al. [41, 42]	67 (15), n = 36	71 (15), n = 28	69 (15), n = 64	SF (evaluate the effect of BMD and bone geometry on the femur strength)	4630 (1550) for men / 3140 (1240) for women
Lang et al. [43]	n = 5	n = 8	73 (11), n = 13	SF/OLS (Assess the proximal femur strength)	574–4640 for SF / 4300–13600 for OLS
Keyak et al. [44]	n = 8	n = 10	70.3[a], n = 18	SF/OLS (identify the femoral fracture load)	573–4640[a] for SF / 3130–15000[a] for OLS
Bouxsein et al. [45]	78 (10), n = 10	82 (13), n = 16	81 (12), n = 26	SF (evaluate the effect of BMD on the femur strength)	3593 (1614) for men / 1997 (1127) for women
Keyak [46]	n = 7	n = 10	70 (52–92)[b], n = 17	SF/OLS (evaluate the effect of loading configuration on the femur strength)	2400[b] for SF / 8500[b] for OLS
Lochmuller et al. [47]	75.8 (11.3), n = 42	81.6 (9.0), n = 63	79.3 (9.9), n = 105	SF/OLS (determine the fracture force)	4230 (1530) for men / 3070 (1060) for women
Eckstein et al. [48]	n = 24	n = 30	79.0 (10.6), n = 54	SF (evaluate the reproducibility error of strength tests)	3951 (1659)
Heini et al. [49]			76.0 (7.2), n = 20	SF (evaluate the effect of femoroplasty-augmentation on femur strength)	2499 (695) for SF / 5764 (1394) for OLS
Manske et al. [50]			69 (16), n = 23	SF (evaluate the correlation between bone geometry and femur strength)	4354 (1886)
Pulkkinen et al. [51]	79.1[b], n = 63	81.7[b], n = 77	81[b], n = 140	SF (identify the geometric determinants of fracture type)	4209 (3477–5468)[a] for men / 2821 (2297–3563)[a] for women
Pulkkinen et al. [52]	78 (11), n = 28	82 (11), n = 34	80 (11), n = 62	SF (evaluate the femur strength from radiographs)	4079 (1165) for men / 2879 (1117) for women

Table 2 (continued)

Study	Mean (SD or range) age in years, sample size			Method (objective of the study)	Mean (SD)/range of fracture force (N)
	Men	Women	Mixed		
Langton et al. [53]	$n = 8$	$n = 10$	52–92[a], $n = 18$	OLS (identify the failure load)	3190–15000[a]
Bakker et al. [54]	85.6 (8.6), $n = 6$	84.8 (6.6), $n = 6$	85.2 (7.2), $n = 12$	SF (evaluate the fracture initiation location)	4032 (370)
Dragomir et al. [55]	63.6 (11.5), $n = 5$	65.8 (9.3), $n = 13$	65.2 (9.7), $n = 18$	SF (identify the fracture load)	2647–5980[a] for men 1409–6179[a] for women
Buijs et al. [56]			65.9 (6.5), $n = 22$	SF (identify the femur strength)	1630–7870[a]
Koivumaki et al. [57, 58]	$n = 20$	$n = 41$	80.4 (9.2), $n = 61$	SF (identify the fracture force)	1520–5460[a]
Nishiyama et al. [59]	70.2 (9.6), $n = 5$	78.9 (13.4), $n = 15$	76.7 (12.9), $n = 20$	SF (identify the fracture force)	1380–4340[a]
Dall'Ara et al. [60]	$n = 17$	$n = 21$	76 (12), $n = 38$	SF/OLS (determine the fracture force)	3800–14200[a] for OLS 1450–5220[a] for SF
Mirzaei et al. [61]	36.5 (13.2), $n = 6$	30.2 (6.7), $n = 4$	34.0 (10.6), $n = 10$	OLS (identify the failure strength)	3430–11600[a]
Gilchrist et al. [62]	82.0 (12.7), $n = 2$	77.2 (9.9), $n = 19$	77.6 (9.9), $n = 21$	SF (evaluate the effect of loading rate on femur strength)	2390 (1410–3720)[a] for impact loading rate 3150 (2170–5300)[a] for fast loading rate 2420 (1460–4440)[a] for slow loading rate
Ariza et al. [63]	73, $n = 1$	76.1 (10.9), $n = 14$	75.9 (10.6), $n = 15$	SF (identify the fracture load by drop tower experiments)	1470–3720[a]
Grassi et al. [64, 65]	40.0 (25.5), $n = 2$		40.0 (25.5), $n = 2$	OLS (identify the failure load)	10619 (3908)
Average across all the studies	73.9	78.5	76.6	–	3462 for SF 7214 for OLS

SF experiments on cadaveric femur under sideways fall loading conditions, *OLS* experiments on cadaveric femur under one-leg stance loading conditions
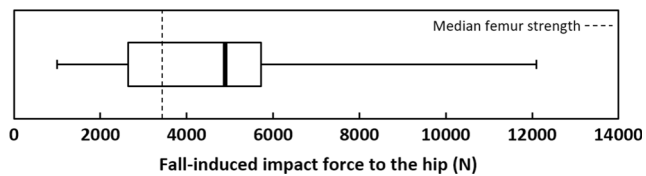
[a] Range is provided

[b] SD is not provided

[c] Elderly subjects

[d] Young subjects

Fig. 1 Box plot showing median, quartiles, and range of lateral fall-induced impact force to the hip in comparison with the median femur strength

experience different impact forces in falls from standing height because of the differences in their physical attributes, T-score predicts the same fracture risk for both. Therefore, this method does not take the effect of impact force into consideration for assessing the risk of hip fracture.

The HSA is constructed based on the principle that a line of pixel values across the bone is summed up in the projected DXA image [69]. Bone quality and bone geometry are the only effective parameters in HSA. Similar to the T-score estimation, HSA does not consider the effect of impact force in assessment of hip fracture risk.

FRAX integrates 12 clinical risk factors to assess the fracture risk for an individual, including age, sex, weight, height, previous fracture, parental hip fracture history, current smoking, glucocorticoids, rheumatoid arthritis, secondary osteoporosis, alcohol consumption, and femoral neck BMD [12, 70]. Although body weight and body height are two of the determinants of impact force and they are indeed included in FRAX, other effective parameters are neglected.

In the FE analysis, introduced to orthopedic biomechanics in 1970s [71], load/constraint conditions are applied to the model, constructed from DXA/QCT image, to calculate the strain and stress distribution in the proximal femur and assess the fracture risk. Compared to other clinical assessment tools, FE analysis considers the effect of impact force more appropriately and this is one of the potential reasons why its accuracy is more convincing [12]. However, many effective parameters are still neglected or not appropriately considered in calculating the impact force, including body segment anthropometric parameters, kinematics of body segments during the fall, nature and direction of the fall, initial conditions of the fall, body configuration before the impact, action of joints, energy absorption during the impact, etc.

It has been demonstrated [12] that in most of the clinical fracture risk assessment tools, bone quality and bone geometry have received much more attention than the impact force. Potential reasons are threefold: (1) difficulties in predicting the subject-specific fall scenario, (2) limitations of the available methods in predicting the impact force, and (3) insufficient investigation of the direct effect of real-life fall-induced impact force on hip fracture risk. In order to facilitate the appropriate consideration of impact force in hip fracture risk assessment tools, there is a need to modify the available impact force predictors. Before that, it should be well-understood what

parameters affect the impact force and how much they differ from subject to subject. The following section evaluates the necessity for developing a subject-specific impact force predictor.

How much impact force differs from subject to subject

Robinovitch et al. [13] illustrated that the peak force in a lateral pelvis impact is a function of (1) impact velocity, (2) effective pelvis mass (i.e., the mass of the pelvis and connecting structures which contribute to the impact loads), and (3) effective pelvis stiffness (i.e., the in-series stiffness of the soft tissue–femur–pelvis complex). All these parameters are interestingly dependent on a complex combination of body parameters [27]. Changes in any of these three factors may significantly affect the magnitude of the applied force to the femur.

Impact velocity

Impact velocity has a positive correlation with body weight and body height. It indicates that heavier and taller subjects experience a higher impact velocity in a fall from standing height. Greater weight and height increases the initial potential energy, which is converted to the kinetic energy at the impact instant and increases the impact velocity and subsequently the impact force. The effects of weight and height on the impact force are even different between men and women. As a result of a biomechanical statistic [27], body weight has a greater effect on the impact force in men than in women. One of the potential reasons for this phenomenon is that the stiffness/damping properties of the body during the impact are differently associated with the STT in men and women. Heavy and tall men not only have thinner hip soft tissues than women, they generally have stiffer tissues [25]. It may be due to the difference in muscle/fat content between men and women (higher muscle content in men compared to women) [13, 72, 73]. Therefore, body anthropometric parameters may significantly affect the impact velocity and impact force.

Cause of fall, that is significantly different in individuals with different age and life style, can also remarkably affect the impact velocity and thus the magnitude of the force applied to the femur. Incorrect weight shifting is the most important cause of fall among the elderly [74] while the cause of fall may be different in younger adults. Differences in the cause of fall affect the initial conditions, i.e., trunk angle and gate velocity (e.g., falls from standing or during walking), and subsequently change the impact velocity. The cause of fall may also affect the duration of the fall that can limit the time for the individual to effectively break the fall with hands or knees. Even if two subjects experience a fall from the same initial

conditions, they may react differently to decrease the severity of the fall, based on their physical abilities, age, and knowledge of effective pre-impact movement strategies. Therefore, impact velocity, as one of the most effective factors on the impact force, can significantly vary from subject to subject.

Effective mass during the impact

Apart from the impact velocity, effective mass, i.e., the mass of that part of the body that moves downwards and contributes to the impact load [13, 14], also changes the magnitude of the force applied to the femur. It is the summation of the effective masses of all body segments at the impact moment [14]. Greater effective mass increases the impact force. In determining the effective mass, the position and orientation of all body segments before the impact are considered in a mathematical model to correlate the vertical position of the hip with its vertical acceleration [75]. Body configuration at the impact instant can thus significantly affect the effective mass. For example, less vertical orientation of the trunk [76] and smaller flexion-extension knee angle (i.e., the angle between the shank and the thigh) [77] can decrease the effective mass and subsequently the impact force. Body segment anthropometric parameters, such as mass, mass center, and mass moment of inertia, are other determinants of the effective mass. Heavier and longer body segments, i.e., shank, thigh, and trunk, have more contribution to the total effective mass [78]. The overall body configuration is another effective factor. Generally, individuals with upside-down pear-shaped body configuration have a greater effective mass during the impact compared to individuals with pear-shaped body configuration [78]. Therefore, the effective mass may also remarkably vary from individual to individual.

Effective pelvis stiffness

The interaction between the body and the ground in the impact stage of a fall is a combination of hyperelastic behavior of the hip soft tissues, fat, skin, muscle, cartilage, and ligaments [79, 80] and visco-elasto-plastic response of the bone [67, 81]. This complex interaction has been mostly simulated by damped vibrational systems, composed of springs and dampers [13]. The spring-damper combination represents the stiffness and damping properties of the soft tissue–femur–pelvis complex that has a significant influence on the amount of absorbed energy during the impact. Table 3 provides a review on the studies that investigated the variation of body stiffness and absorbed energy among different individuals.

As shown in Table 3, body stiffness and damping coefficient remarkably varies among different individuals since it is affected by trochanteric soft tissue thickness (STT), BMI, body configuration at the impact instant, landing surface, condition of muscles (relaxed/contracted), etc. Among all

parameters, body mass index, trochanteric soft tissue thickness, and muscle conditions are the most effective factors on the body stiffness during the impact. It has to be pointed out that the effective pelvis stiffness not only depends on the body parameters but also on the force-deflection and force-velocity properties of the pelvis and the ligaments connecting the pelvis to the trunk and thigh [13, 18]. Age [90] and sex [27] also affect the pelvis stiffness and subsequently the hip fracture risk (Fig. 2). For example, if a man and a woman in similar range of age with the same trochanteric soft tissue thickness and effective mass experience a lateral fall with the same impact velocity, the impact force for the male subject is generally higher than that for the female subject [25, 27]. It indicates that the amount of absorbed energy during the impact is affected by sex due in part to the differences in the muscle/fat content in the hip region. In addition to the mentioned parameters, floor surface may also change the magnitude of the impact force [89]. The floor can be hard wood or carpeted that affects the amount of absorbed energy during the impact, and thus, people with different life styles may experience different impact forces in falls on the hip.

Since the body stiffness is one of the most dominant parameters affecting the pelvis stiffness and subsequently the hip fracture risk, many studies have devoted to wearable protective pads [66, 91–93]. Hip protectors significantly contribute in attenuating the impact energy and may reduce the axial compressive force induced to the femur by 20% [94] and the peak pressure by 70% [95], especially for low BMI subjects who generally do not have thick trochanteric soft tissues. Research studies have thus shown that these pads can remarkably reduce the risk of fall-induced hip fracture.

In general, impact velocity, effective mass, and effective stiffness as determinants of the hip impact force are influenced by age, sex, body anthropometric parameters, body configuration at the impact instant, pre-impact movements, hip soft tissue thickness, and muscle conditions, and thus, the impact force significantly varies from subject to subjects. Since the impact force may remarkably affect the risk of hip fracture, accurate subject-specific prediction of impact force can be considered as of great significance in improving hip fracture risk assessment tools.

Available models to predict the impact force

The available models in the literature to simulate the body falling are generally composed of two parts: (a) a dynamic model to simulate the kinematics of body segments during the fall and (b) an impact model to simulate the interaction between the body and the ground. In this section, a review is provided on the available dynamic and impact models for simulating the body falling and their strength/weaknesses are investigated.

Table 3 Variation of body stiffness in different subjects: how attenuated force during the impact is affected by body parameters and varies from fall to fall

Study	Method and study population	Results
Robinovitch et al. [13]	Pelvis-release experiments on 14 human subjects to measure the body stiffness and damping under muscle-relaxed/contracted conditions	- Increased trochanteric soft tissue thickness and impacting the ground in a relaxed state can both decrease the effective body stiffness and subsequently reduce the impact force
Malmivaara et al. [82]	Study on 19,518 adults with the age between 20 and 92 years to determine the association between the fall-induced fracture risk and the body mass index (BMI)	- For women aged <45 years, BMI is associated with an increased fracture risk - Greater BMI is a protective factor among both men and women aged ≥65 years
Greenspan et al. [83]	Study on 149 fallers (126 women, 23 men) with the age of 65 years and older, including 72 fallers with hip fracture and 77 fallers with no fracture, to identify the effect of body parameters on hip fracture risk	- Reduced STT allows more energy to be transmitted to the femur - Low BMI elderly are more prone to hip fracture
Robinovitch et al. [15]	Using trochanteric soft tissues harvested from the cadavers of nine elderly individuals in impact pendulum experiments, to investigate why BMI is a protective factor against hip fracture	- STT is an influential factor in absorbing the impact energy in a lateral fall - Less STT in low BMI subjects is the potential reason why they are more prone to hip fracture
Hayes et al. [17]	Fall experiments from a height of 0.7 m	- Body stiffness and applied force to the femur in the muscle-active state is remarkably higher than the muscle-relaxed state
Laet et al. [84]	A meta-analysis on 60,000 men and women to quantify the effect of BMI on hip fracture risk	- Low BMI persons are in a greater risk of hip fracture - Obesity should not be considered as a protective factor for hip fracture; rather, leanness is a significant risk factor
Bouxsein et al. [85]	Investigating the effect of STT on hip fracture risk by a study on 49 women, where 21 individuals experienced a hip fracture in the 13-year follow-up period	- Trochanteric soft tissue thickness influences hip fracture risk by attenuating the impact energy - Each 16.7 mm decrease in trochanteric soft tissue thickness is associated with a 1.8-fold increase in hip fracture risk
Majumder et al. [80]	Modeling of pelvis-femur-soft tissue complex to determine the effect of trochanteric soft tissue thickness on the risk of hip fracture in sideways falls	- Increase in trochanteric soft tissue thickness would non-linearly reduce the risk of hip fractures - By 81% decrease in hip soft tissue thickness, peak impact force increases by 38%
Armstrong et al. [86]	Study among 925,345 women, followed for an average of 6.2 years, to determine the effect of physical activities and BMI on hip fracture risk	- Increase in fracture risk per unit decrease in BMI is significantly greater among thin women than among obese women - Thin women are relatively in a higher risk of hip fracture - Overweight and obese women are at lower risk of hip fracture
Johansson et al. [87]	A meta-analysis on 398,610 women with an average age of 63 years (range 20–105) from more than 25 countries to investigate the effect of BMI on fracture risk at different skeletal sites	- Low BMI is a risk factor for hip fracture; however, it is a protective factor for lower leg fracture - High BMI is a protective factor for osteoporotic fractures - High BMI is not as strong protective factor as low BMI is a risk factor
Levine et al. [25]	Pelvis-release experiments on 28 participants (mean (SD) age=22.4 ± 2.7 years) to identify the effect of BMI on effective pelvis stiffness	- High BMI subjects have higher effective mass and also higher capacity of absorbed energy - High BMI subjects have higher pelvic landing surface and thicker soft tissues on the hip, but the maximum force applied to the femur in a fall is still higher in them

Table 3 (continued)

Study	Method and study population	Results
Majumder et al. [88]	Modeling the pelvis-femur-soft tissue complex to obtain the most effective criteria for the hip fracture among anthropometric parameters	- Trochanteric soft tissue thickness is the most dominant criterion for hip fracture among all anthropometric parameters - A tall and heavy person with thicker trochanteric soft tissues may escape hip fracture in lateral falls
Bhan et al. [89]	Lateral pelvis-release experiments on 20 university-aged participants (10 men and 10 women) to characterize the effect of BMI on the energy absorption during sideways falls on the hip	- Energy absorption may explain the reason for the reduced risk of hip fracture for persons with high BMI - High BMI persons absorb more than threefold greater energy than low BMI subjects
Choi et al. [90]	Determining hip soft tissue stiffness and damping of 34 women (17 old and 17 young) to investigate the effect of age on tissue properties	- Stiffness and damping of hip soft tissues reduce by aging - Dramatic rise in the number of hip fractures with age is associated with the reduced capacity of hip soft tissues to absorb and dissipate energy
Nasiri and Luo [27]	Investigating sex-differences in the association between hip fracture risk and body parameters in a study population of 130 subjects (50 men and 80 women)	- Increased BMI is a protective factor against hip fracture in women, but not in men - Low BMI is a risk factor for hip fracture in both men and women
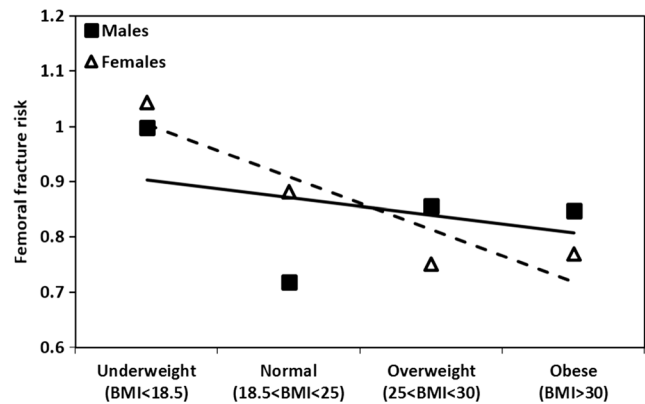
Fig. 2 Femoral fracture risk at various levels of BMI (kg/m^2) for 50 men and 80 women (reproduced from [27] with permission). Fracture risk increases with decreased BMI (but with different trends in men and women) due to the lower capacity of hip soft tissues to absorb impact energy

Human body dynamic models for simulating the fall

The purpose of dynamic models, presented in the literature, is to predict the hip impact velocity in a fall. The simplest yet effective biodynamic models for simulating the lateral fall are point-mass (Fig. 3a) and one-link single-degree of freedom (DoF) (Fig. 3b) models based on energy conservation [14]. The hip impact velocity is calculated by equating the available potential energy at the onset of the fall to the potential and kinetic energy at the impact moment. Assuming the location of the hip at the middle of the body, the hip impact velocity by the point-mass and the one-link model are respectively represented by Eqs. (2) and (3), where h_{cg} is the height of the centre of gravity of the body.

$$V = 4.43\sqrt{h_{cg}} \tag{2}$$

$$V = 3.84\sqrt{h_{cg}} \tag{3}$$

Both models have a single DOF and the only difference is that in the point-mass model the mass falls vertically while the one-link model includes the effect of rotational energy. Equations (2) and (3) show that by a slight change in the assumptions, the resulted velocities are considerably different.

A two-link model has also been proposed to represent lower extremity (leg segment) and upper extremity (trunk), interconnected by a hinge at the hip joint [14]. The trunk orientation may vary prior to impact, and thus, two different configurations are considered for the trunk angle, i.e., vertical (Fig. 3c) and 45° (Fig. 3d). The hip impact velocity in vertical and 45° configuration of the trunk is respectively represented by Eqs. (4) and (5), where h is the total body height.

$$V = 3.24\sqrt{h} \tag{4}$$

$$V = 2.72\sqrt{h} \tag{5}$$

Fig. 3 **a** Point-mass, **b** single-link, **c** vertical trunk, and **d** 45° trunk model for simulating the body falling from standing height [14]. **e** Projection of three-link human body dynamic model on sagittal and coronal plane (reproduced from [77] with permission). **f** Right posterolateral view showing the 11-link dynamic model to simulate sideways body falling (reproduced from [99] with permission)
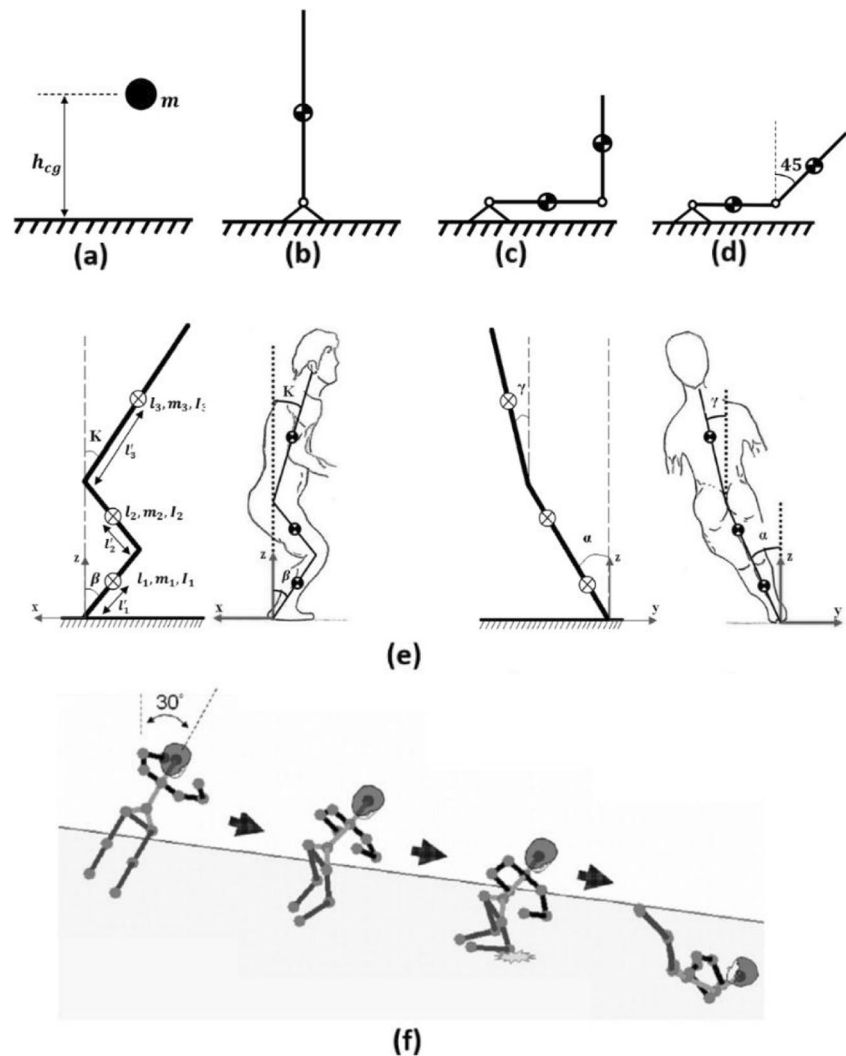
Inspired by a pioneering work performed by Kroonenberg et al. [14], a three-link model is also proposed [77, 78, 96] to simulate the motion of shank, thigh, and trunk in a fall from sideways (Fig. 3e). In this model, the body is represented as an open chain of three links connected by frictionless hinges [14]. This model has two rotational degrees of freedom for the hip, i.e., flexion/extension and abduction/adduction, one flexion/extension degree of freedom for the knee, and an abduction/adduction degree of freedom for the ankle. Rotational springs are incorporated in the hinges to account for muscle forces acting on the joints. Required body segment anthropometric parameters, including length, mass, mass center, and mass moment of inertia, are derived from the whole-body DXA image of the subject [77, 97] and are assigned to the corresponding links. Inverse-dynamic approach was used to take the effect of hip and knee joint torques into account [98]. Subject-specific consideration of body segment anthropometric parameters and joint torques improved the accuracy of the model in predicting the hip impact velocity [77, 78].

More complex models have also been introduced to simulate the fall from sideways and determine the impact velocity. Lo and Ashton-Miller [99] proposed an 11-link model, including shanks, thighs, pelvis, trunk, head, upper arms, and forearms, to simulate the lateral fall from standing height (Fig. 3f). The rigid segments were connected by frictionless hinges (knees, neck, and elbows), universal (waist and shoulder), and ball-and-socket joints (hips). The rotational joint torques were simulated by a pair of agonist and antagonist joint muscle torque actuators. Table 4 shows the range of error for available models in predicting the hip impact velocity in a fall from sideways.

Comparison between the error of different models in Table 4 shows that increasing the complexity of the model does not necessarily result in improved predictions. For example, the accuracy of the two-link model (with the 45° jackknife configuration) in predicting the impact velocity is greater than that of the three-link model (A) [14]. Therefore, increasing the number of links without a proper consideration of the joint

Table 4   Range of error in predicting the hip impact velocity in sideways falls

Method	Relative error in predicting the hip impact velocity in sideways falls
Point-mass one degree of freedom [13, 14]	42 % <error<58%
One-link one degree of freedom [14]	23 % <error<37%
Two-link model (vertical jackknife) [14]	45 % <error<62%
Two-link model (45° jackknife) [14]	22 % <error<36%
Three-link model (A) without taking the joint torques into account [14]	48 % <error<61%
Three-link model (B) with consideration of the muscle torques in the knee joint [14]	34 % <error<46%
Three-link model (C) with consideration of the subject-specific anthropometric parameters and muscle torques in hip and knee joint [77, 78]	3 % <error<10%
11-link model with joint torques at knees, hips, shoulders, neck, elbows, and waist [99]	2 % <error<27%

torques may even increase the uncertainties of the model and thus reduces the accuracy of predictions. The error of the three-link model (C) (3–10%) [78] is lower than that of three-link model (A) and (B) that indicates the significance of subject-specific consideration of effective parameters in modeling. However, the three-link model (C) is validated by only three subjects due to the difficult nature of the performed fall experiments. Further validations for this model can improve its application in clinical studies. The error of the 11-link model that simulates the kinematics of all body segments [99] is 2–27% in predicting the hip impact velocity, which seems to be the most accurate model among the available ones. Therefore, the number of links and consideration of joint torques and subject-specific body segment anthropometric parameters are the factors that together can effectively improve the ability of the model in predicting the hip impact velocity in a fall. Although increasing the number of links with proper consideration of joint torques improves the accuracy of predictions, it also increases the complexity of the model and subsequently the computational time and storage. Since the models are being used in clinical applications, a combination of model complexity, model accuracy, computational efficiency, and ease of adopting in clinics should be considered to select an appropriate dynamic model for hip fracture risk assessment tools.

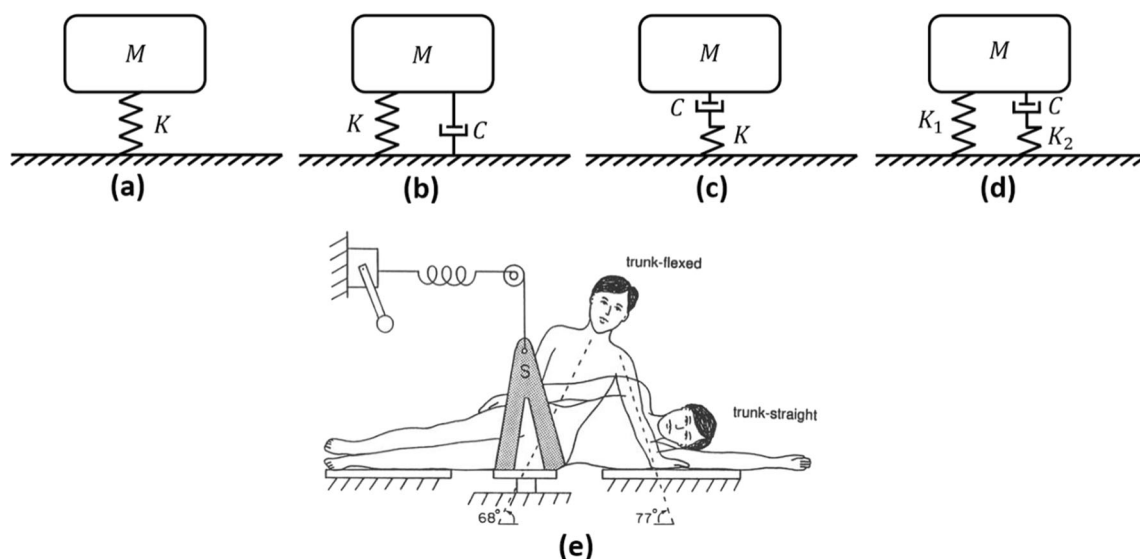## Models to simulate the impact stage of a fall and determine the impact force

A benchmark work by Robinovitch et al. [13] simulated the interaction between the body and the ground by a single-degree of freedom damped vibrational system as shown in Fig. 4a. m is the effective mass and K and b are, respectively, body stiffness and damping properties, that are determined from force-deflection and force-velocity properties of the trochanteric soft tissues, muscles of the pelvis, abdomen, lower legs, and the skeletal components [13]. This is the simplest effective model capable of simulating the oscillatory response of the body during the impact stage of a fall.

Robinovitch and colleagues also proposed three nonlinear single-degree of freedom (DOF) shock-absorbing systems for simulating the impact stage of a fall [19], i.e., Voigt support model (Fig. 4b), Maxwell support model (Fig. 4c), and standard linear solid support model (Fig. 4d). Results showed that the Maxwell and standard linear solid support models are better predictors of the peak impact force. Trunk-straight and trunk-flexed configurations of the hip during the impact were also simulated by a two DOF model (Fig. 4e) in order to determine the distribution of contact force to the hip [18]. This model considered both compressive and flexural motion of the body during the impact where both compressive stiffness and damping coefficients were significantly greater than flexural ones. Results showed that with the flexed trunk, the stiffness and the applied force to the hip are significantly increased. The increase in the stiffness is due to the changes in the geometrical relations between femur, pelvis, and supporting ligaments [18]. Also, in the trunk-upright configuration, a larger trunk mass is located closer to the hip that remarkably increases the effective mass during the fall [18]. Although the implemented damper in the vibrational models is to take the effect of damped energy into account, experimental studies showed that the proximal femur region absorbs the vast majority of the impact energy, and only 15% of the total impact force is applied to structures surrounding the hip [18].

Force-deflection properties of the pelvis are also measured in vivo by Robinovitch et al. [13, 18] and Laing et al. [24] to evaluate whether characterization of nonlinearities in stiffness affects the accuracy of a mass-spring model in predicting the peak impact force. Results demonstrated that the force-deflection is slightly nonlinear (limited to force below 300 N), characterized by increasing stiffness as deflection increases. However, for loads beyond 300 N stiffness remained constant.

The characteristics of the introduced impact models are determined from pelvis-release experiments with living humans [13, 17, 18, 24, 25, 28, 89, 90, 94, 95], cadaver-

Fig. 4 **a** Mass-spring model, **b** Voigt support model, **c** Maxwell support model, and **d** standard linear solid support model for simulating the impact stage of a fall. **e** Trunk-straight and trunk-flexed configuration in pelvis-release experiments (reproduced from [18] with permission)

based studies [15], or surrogate human pelvis-pendulum-foam soft tissue [19, 66]. Each approach has its own advantages and limitations. The chief strength of the cadaver-based studies is that the cadaver can be subjected to the impact force occurring in real-life falls without any limitation or simplification in performing the experiments. However, the response of the cadaver may considerably differ from that of the living human, due to the post-mortem changes in the skin and fat in the hip region, lack of active/passive joint muscle actions, and lack of cognition to avoid the fall by real-life pre-impact movements. Furthermore, kinematic constraints are required in order to mimic a real-life fall by a cadaver that will affect the dynamic response [18]. The main weakness of using living humans is that the experiments can only be performed at safe loading levels, which is approximately 250 N [18]. The potential issue is that trochanteric soft tissues show nonlinear force-deflection and force-velocity properties only at low force region (0–300 N) [24] while the response of body at the low force regime is not the representative of that occurring

in an actual fall [18]. Furthermore, experiments are only conducted on young adults and the results may not be interpretable for the elderly. For experiments on foam as a surrogate of human soft tissues, again results may significantly differ from what is happening in real-life falls, due to the remarkable differences between the structure of foam and human soft tissues. The other limitation for pendulum experiments with femur covered by foam is that it only simulates the effect of trochanteric soft tissues in attenuating the impact energy, while in a real-life fall, energy absorption during the impact is a complicated mechanism where other structures also contribute, including pelvis, muscle and ligaments that connect the pelvis to the trunk and lower limbs, and femur [18].

In addition to the aforementioned models, Lo and Ashton-Miller proposed formulas (Eqs. (6) and (7)) to separately calculate the stiffness and damping of soft tissues and boney parts for body segments that could potentially impact the ground, such as heels, knees, greater trochanters, pelvis, shoulders, thorax, elbows, wrists, and head [99, 100].

$$\begin{cases} K = 3.5 \left(\mathrm{MN}\big/\mathrm{m}^3 \right) \\ B = 6 \left(\mathrm{m}\big/\mathrm{s} \right)^{-1} \end{cases}, \text{ for } \textit{soft tissue impact over the lateral, anterior and posterior surfaces of the pelvis, thorax, and shoulders.} \tag{6}$$

$$\begin{cases} K = 10.0 \left(\mathrm{MN}\big/\mathrm{m}^3 \right) \\ B = 7 \left(\mathrm{m}\big/\mathrm{s} \right)^{-1} \end{cases}, \text{ for } \textit{boney tissue impact at the heel, knee, greater trochanter, elbow, wrist, and head.} \tag{7}$$

Although specific determination of the stiffness and damping properties of each body part during the impact can remarkably increase the accuracy of predictions, these

properties are not the same for all subjects. As an instance, the stiffness and damping of the anterior and posterior surfaces of the pelvis, thorax, and shoulders can considerably differ

between overweight and underweight subjects and also between men and women [27], while Eqs. (6) and (7) consider the same stiffness and damping for all individuals.

Apart from the introduced vibrational models to simulate the impact stage of a fall, two empirical equations are also proposed to predict the peak impact force in a lateral fall. Equation (8) is proposed by Kroonenberg et al. [14], where $n = V \frac{\sqrt{\left(\frac{71000}{EM}\right)}}{9.81}$, $V = 2.72\sqrt{h}$, $\omega = \pi - \tan^{-1} n$, m_e is the effective mass (0.35 of body weight), and V is the impact velocity.

$$F_{peak} = m_e \times g \times (n\sin(\omega) - \cos(\omega) + 1) \quad (8)$$

Equation (9) is obtained by Robinovitch et al. [13] and extrapolated by Yoshikawa et al. [101], where h is body height in centimeters and W is body weight in Newton.

$$F_{peak} = 8.25 \times W \times \sqrt{\left(\frac{h}{170}\right)} \quad (9)$$

The effective parameters in both empirical equations are body weight and body height. Ignoring other effective parameters, such as trochanteric soft tissue thickness, influences the accuracy of the equations in predicting the peak impact force [77].

A method is also proposed by Robinovitch et al. [15] to consider the effect of hip soft tissues in attenuating the force applied to the greater trochanter. They suggested that for each millimeter increase in tissue thickness, peak force at the greater trochanter level decreases by 71 N. In most FE analyses of the proximal femur, a combination of Kroonenberg and Robinovitch method is implemented to predict the force applied to the femur in a sideways fall as follows:

$$F_{peak} = m_e \times g \times (n\sin(\omega) - \cos(\omega) + 1) - 71 \times STT \quad (10)$$

Although it is a simplification to combine Eq. 8 with Robinovitch method, as in study of Robinovitch et al. [15], the difference between the forces at soft tissue surface and at the greater trochanter was not reported to be 71 N/mm; however, Eq. 10 is the most common method for determining the fall-induced impact force in hip fracture risk assessment tools. Results of hip fracture studies, particularly FE analyses, proved the effectiveness of this formula in estimating the fall-induced impact force to the hip.

Furthermore, a subject-specific damped vibrational model has been proposed [77] to simulate the interaction between the body and the ground. The stiffness and damping properties of the impact model are derived subject-specifically considering the gender of the individual, as well as the thickness of trochanteric soft tissues [27]. However, several simplifications are still used in modeling, including the limited number of degrees of freedom and neglecting the risk of fall and pre-impact movement strategies. Whole-body DXA image is also needed by this model that may affect its applicability in clinics.

Missing/neglected variables in available models for predicting fall-induced impact force

A fall has three phases: (1) the instability phase, (2) the descent phase, and (3) the impact phase [17, 102]. There are many variables in each of these phases that may significantly change the kinematics of the fall and subsequently the impact force. In this section, missing parameters in currently available models for predicting the impact force are investigated and suggestions for future studies are provided.

Instability phase

Instability phase is directly affected by the cause of fall. Most of the models are focused on determining the mechanics of fall without considering how and why the fall occurs. The common causes of falls among the elderly are incorrect weight shifting (41%), trip or stumble (21%), hit or bump (11%), loss of support (11%), collapse (11%), and slipping (3%) [74].

Elderly people mostly initiate the fall because of incorrect transfer or shifting of body weight. As an instance, elderly with walkers who are standing and want to initiate a turn by rotating the walker are likely to lose their balance as the center of gravity is moved outside the base of support [74]. Trip and stumble are mostly attributed to foot catching on the ground and difficulty in raising the foot [74]. Collisions with environmental objects and being pushed or pulled by another person are the most common causes of falls by hit or bump [74]. The cause of fall affects the initial conditions of the faller, the direction of the fall, i.e., sideways, forward, and backward, the kinematics of body segments during descent phase, possibility of using pre-impact movement strategies, and thus the magnitude of the impact force. As an instance, the possibility of stepping, squatting, forward walking, sitting down, and use of knees to break the fall is different in falls from different directions. Also, a fall due to trip, stumble, or slipping may cause a greater impact force compared to a fall due to incorrect weight shifting as the initial velocity at the moment of loss of balance is greater and the time to effectively arrest the fall is shorter. Common activities at the time of fall initiation, i.e., walking forward, standing quietly, sitting down, and initiation of walking [74], and gate speed also affect the instability phase. It has been demonstrated that slips and faints lead to forward falls at fast gate speed. However, at lower gate speed, slips and faints result in sideways falls with impact on the hip, suggesting a greater risk for hip fracture [103]. Not only the cause of fall varies among different subjects due to the differences in the life style and thus environmental hazards but also one individual may experience falls with different initial

conditions. Therefore, if not impossible, it is very difficult to consider the subject-specific cause of fall for hip fracture risk assessment purposes. That is the potential reason why the cause of fall is totally neglected in hip fracture risk assessment tools. Determining the age-based most frequent initial conditions using database of video-captured falls [74] may be implementable for the dynamic models to predict the impact force for a subject with a specific age. Another reasonable way is to consider the most common cause of fall among the elderly (incorrect weight shifting [74]) and extract the corresponding initial conditions to be used for prediction of the impact force in hip fracture risk assessment tools. Also, an investigation can be conducted to find the most dangerous and sever cause of fall among the elderly to be adopted in impact force predictions. So far, results of typical falls of young adults on a thick foam mattress have been used for determining the initial conditions of a fall from sideways [14, 16, 23]. However, findings of video capture of the circumstances of falls in the elderly people [74] can be more useful in extracting the initial conditions of a real-life fall. Improving estimation of the initial conditions based on the most frequent cause of fall may increase the accuracy of predicted impact force and hip fracture risk.

Descent phase

Descent phase is between the instant of instability and the impact moment. The most important parameters in this phase that change the magnitude of the impact force are the action of joints and the use of body segments to arrest the fall, i.e., pre-impact movement strategies. Rotational springs [14] and time-dependent functions [78, 98] have been proposed to be accounted for the action of hip and knee during descent. However, one of the criticisms of the currently available methods is that the action of joints is not considered subject-specifically while joint actions may significantly vary from subject to subject. Sedentary individuals may less properly use their joints to decrease the severity of the fall compared to athletes. The joints action during the descent phase may also be culture-dependent. In many Asian countries, people frequently adopt floor sitting in their daily living activities [104, 105], and thus, they may be more experienced in effectively using their body segments to control their balance by applying appropriate torques in their joints. This experience may help them to more appropriately react during the descent phase. The age is another effective parameter, i.e., younger subjects may be able to more properly use their joints to decrease the severity of the fall [23].

Pre-impact movements can also remarkably influence the hip impact force. Although any fall from standing height can potentially lead to a hip fracture, only less than 2% of falls in older adults result in hip fracture [106], suggesting that protective responses can effectively reduce the severity of the fall

and facilitate safe landing. Breaking the fall using outstretched hands and stepping are the most common strategies to reduce the impact force in young adults [23, 107]. These strategies can be more effectively implemented in sideways and forward falls [21, 100, 108, 109]. Impacting with knees [23], forward/backward rotation [21], squatting [22], eccentric contraction of lower extremity muscles [20], and knee extension during the final stage of descent [20] are other strategies to reduce the severity of the fall. It has been demonstrated that the hip impact force can be reduced by 56% if the subject uses the knee, hip, and spine rotations to arrest a lateral fall [99]. The reasons for this force reduction include distribution of the impact over a large body area and allowing more time for dissipation of the impact energy [99]. Experimental studies showed that martial arts, such as changing the fall into a rolling movement, can also reduce the hip impact force by 27 – 30% [76]. Holding an object to break the fall is another common activity that may significantly change the kinematics of fall [29] and subsequently reduce the applied force to the hip. Not only the nature of the protective responses affects the magnitude of the impact force but reaction time also influences their effectiveness. Increase in reaction time among the elderly may explain why many hip fractures occur among the older adults but few occur among younger people [107]. Table 5 provides a review on how pre-impact movements can change the magnitude of the impact force.

Although using body segments to break the fall is quite common as an inherent activity, pre-impact movement strategies have not been appropriately considered in available models for predicting the impact force. Concluded from Table 5, the action of an individual to arrest the fall may significantly differ from fall to fall based on many parameters, including the cause and initial conditions of fall that affect the duration of the descent phase, environmental parameters such as the darkness of the room that affects the implementation of the pre-impact movement strategies, subject's physical ability, subject's knowledge of pre-impact movement strategies, and the level of cognition of the faller during the descent phase. Therefore, it is impossible to consider the subject-specific pre-impact movements in dynamic models for determining the impact force. Again, the applicable method is to determine the most frequent reaction of the elderly during the descent phase to be considered in modeling. As initiated by Choi, Wakeling, and Robinovitch [116], the valuable recorded videos of real-life falls [74] can be used to identify common pre-impact movements to arrest the fall. Then, mathematical modeling of the pre-impact movements can be added to the impact force predictors for more reliable hip fracture risk assessment. Although hip fracture risk assessment tools are being used in clinics for individuals with varying range of age, the elderly are mostly prone to hip fracture due to the higher risk of fall, lower chance to effectively break the fall, and lower BMD. It may help explain why this study recommends

Table 5 Effect of pre-impact movement strategies and martial art techniques on the reduced impact force to the hip

Study	Method and study population	Results
Nevitt et al. [110]	Investigation among 325 subjects over age 60 (266 women and 59 men) who had fallen at least once	- Slow (compared to fast) hand reaction may increase the risk of hip fracture as it cannot effectively decrease the applied impact force
Hsiao and Robinovitch [106]	Abrupt fall experiments without warning on gymnasium mattress by three men and three women, ranging in age from 22 to 35 years, to investigate the effective movement strategies for preventing fall-induced injury	- Impacting with wrist at almost the same instant as the pelvis will share the contact energy between the two of them - Trunk rotation in a sideways fall helps to prevent the impact to the lateral aspect of the hip - A repeatable series of body segment movements which facilitate safe landing may reduce the impact force
Sabick et al. [107]	Fall experiments from kneeling position on seven men and two women, aged 25–55, to evaluate active responses to decrease the impact force in sideways falls	- Impact forces at the hip and the shoulder were 12 and 16% less for falls with attempting to break using hands compared to falls without attempting to arrest - Active pre-impact responses reduce the impact forces experienced at the hip and the shoulder in lateral falls
Robinovitch et al. [111]	Fall experiments on four men and five women, aged 19–36, to investigate the effect of body configuration at the impact moment on the applied force to the femur	- Falls with large horizontal excursions of the center of gravity with respect to the ankle may increase the impact velocity and kinetic energy
Sandler and Robinovitch [20]	Developing three inverted pendulum models of fall from standing height to investigate the effective methods for reducing the body impact velocity and kinetic energy during the impact	- Eccentric contraction of lower extremity muscles during descent leads to energy absorption by up to 150 J - Generation of flexion rotations and eccentric extensor torques at the knees and hips during descent reduces the severity of the fall - Knee extension at the descent phase of a fall transfers up to 90 J of impact energy into horizontal (as opposed to vertical) kinetic energy
Robinovitch et al. [21]	Sideways fall experiments on 22 women, ranged in age between 18 and 35 years, to test the effect of forward and backward rotation on the reduced impact force	- Forward/backward rotation during descent can help the faller to avoid impact to the hip and lower the risk for hip fracture - Impacting with knees and outstretched hands prior to hip can remarkably reduce the impact force
Robinovitch et al. [22]	Fall experiment on 22 women, aged 21–29, to test the effect of squatting during descent on the reduced impact force	- Squatting during descent decreases the impact severity - Delay in squatting during descent reduces its protective value
Weerdesteyn et al. [112]	Investigation on 113 elderly, aged 73 ± 4, with the history of fall to evaluate whether elderly are able to learn martial art fall techniques	- Training martial art fall techniques can help the elderly to reduce the number of fall, be successful in obstacle avoidance, and increase the balance confidence
Groen et al. [76]	Sideways fall experiments from kneeling on 6 experienced judokas (age 24.2 ± 3.8) to get insight into the role of martial art techniques in reducing the impact force	- Martial art fall techniques can reduce the impact force by more than 25% - Rolling after impact and less vertical trunk orientation are of effective martial art fall techniques
Feldman and Robinovitch [23]	Sideways fall experiments on young adults (13 men and 31 women, aged 19–26) to evaluate the effect of impact to hands and stepping on the applied force to the femur	- For each 10 ms increase in the interval between hand and pelvis impact, the pelvis impact velocity decreases by 3.6%. Therefore, the faster the subject breaks the fall with hands, the smaller impact force is induced to the femur - Stepping prior to impact can reduce the impact force by 22% on average

Table 5 (continued)

Study	Method and study population	Results
		- Effective use of hands and stepping can interpret the lack of hip fracture in young adults - Even unsuccessful attempts to recover balance during a fall by stepping may reduce the risk for hip fracture because it increases the time to pelvis impact, increases the time between hand and pelvis impact, and decreases the pelvis impact velocity
Weerdesteyn et al. [113]	Investigating whether hip impact force in martial art lateral falls is smaller than in a natural lateral fall arrest strategy (naturally blocking the fall with outstretched arms) in young adults (10 women, aged 28.3 ± 6.6) after a 30-min training session	- Martial art fall techniques, achievable in a very short training time, can substantially reduce the impact force - Training martial art techniques is a very promising method to be used in interventions to prevent fall injuries
Lo and Ashton-Miller [99]	Developing a 3D human body dynamic model with 11 segments and controller on joints to match actual and desired joint angles in a lateral fall and investigate the effect of pre-impact movement strategies on the hip impact force	- Distributing the impact over a large body area and allowing more time for dissipation of the impact energy can reduce the impact force - Using the lateral aspect of the knee as the first impact site can dissipate more energy before hip impact - Flexion of the lower extremities, ground impact with the side of the knee, and axial rotation to present the posterolateral aspect of the thigh, pelvis, and then torso can reduce the hip impact force by 56%
Groen et al. [114]	Evaluating the effect of learning martial art fall techniques among elderly (6 men and 19 women, aged 60–81) on their performance during a sideways fall	- Healthy older individuals can learn the basics of martial art fall techniques within five weekly training sessions - Learning martial art techniques reduces the fear of falling that may result in the prevention of falls
Zijden et al. [115]	Sideways fall experiments on 12 experienced judokas (age: 27.6±10.7) to compare the effect of martial art techniques and block techniques on the hip impact force	- Martial art techniques, such as converting the fall to a rolling movement, can reduce the hip impact force by 27% compared with block techniques (blocking the fall with outstretched arm) - Martial art techniques can change the magnitude of the hip impact force but do not change the loading configuration

using the elderly database for improving the impact force predictors.

From Table 5, not only martial art fall techniques can substantially reduce the impact force but also learning them may reduce the fear of falling that can result in prevention of falls. Surprisingly, learning martial art fall techniques is achievable in a very short time [113], even for the elderly people [114]. It has been demonstrated that older individuals can learn the basics of protective fall techniques within five weekly training sessions [114]. Nevertheless, instructing/learning fall techniques has not received much attention since its importance has not been emphasized. Future studies can invest on introducing new martial art fall techniques and propose guidelines for vastly implementing them among the elderly.

Impact phase

The last phase in a human body falling is when the hip impacts the ground. Fall direction that affects the boundary conditions of the proximal femur in the impact phase is an important parameter in simulating the proximal femur under fall-induced impact force. As an instance, in an impact with the lateral aspect of the hip, the force is applied to the greater trochanter while in a forward/backward fall the force is applied to the anterior/posterior aspect of the femur. Evidence showed that the impact during the fall to the sides, as opposed to the anterior/posterior aspect of the pelvis, substantially increases the risk for hip fracture [21]. Surprisingly, more than 60% of hip fractures are caused by sideways falls [5, 9], although the number of forward falls is three times that of lateral falls [1, 117]. The magnitude of the impact force is also influenced by the impact site. Potential reasons are the difference in the thickness of soft tissues overlying the hip in lateral, anterior, and posterior aspects of the hip and the difference in boundary conditions of the femur during the impact. Other effective parameters in the impact phase are the condition of muscles (relaxed/contracted), the portion of muscles and ligaments that contribute to the impact which is depended on the landing surface, abdominal and thoracic viscera, flexural deformation of the spine, and muscular connections between the trunk and the thigh [18]. The hardness of floor is also an effective parameter in the impact stage. Falling on a hard floor and a carpet does not lead to a same impact force. Again, it is not applicable to consider the impact site, muscle conditions, and floor hardness, subject-specifically, as these parameters are not constant for an individual. Therefore, lateral fall on a hard floor as the most sever scenario can be considered in impact force predictors. Also, the structure of muscles and ligaments peripheral to the hip and the joint torques that the individual applies during the impact is very complex to be taken subject-specifically into account for determining the impact force. There is a controversy between the effect of muscle conditions and pre-impact movements to reduce the severity

of the fall. Although results of fall experiments have shown that impact with relaxed muscle will cause a safer landing [13, 17], any reaction in the lower extremity to arrest the fall may contract the muscles peripheral to the hip. Therefore, a comprehensive research is in need to determine how the muscle conditions and pre-impact movements should compromise to reduce the impact force.

One of the criticisms of the available models in predicting the impact force is that only the thickness of hip soft tissues has been considered for determining the attenuated impact energy. Studies on impact stage of human body falling have shown that a complex combination of interdependent body parameters can also influence the attenuated energy, including the fat and muscle content, age- and sex-related properties of soft tissues, flexural deformation of the spine, connection between trunk and lower limb, muscle conditions (relaxed/contracted), and landing surface [13, 15, 18, 24, 27, 78, 80, 90]. Therefore, future studies are still in need to determine the subject-specific stiffness/damping properties of the body for accurate prediction of fall-induced impact force.

Furthermore, research studies have mostly been focused on determining the peak impact force applied to the femur in a sideways fall. Since our understanding of the visco-elasto-plastic properties of the bone has recently been improved [38, 67, 68, 81, 118–120], prediction of the impact force time history may significantly improve the accuracy of future FE models in hip fracture risk assessment. Although considerable amount of studies have devoted to determine the strain-dependent properties of bone, these properties have not been adequately implemented in hip fracture risk assessment tools. Predicting the time history of impact force is much more helpful when the strain-dependent bone failure material properties are well investigated.

Conclusions

Sideways fall-induced impact force is one of the important determinants of hip fracture risk. Nevertheless, the available methods have limitations in predicting the impact force. These limitations are mostly due to the great number of uncertainties in simulating an unexpected real fall, including fall direction (forward, backward, or sideways), initial conditions of fall, impact site (anterior/posterior side of the femur or greater trochanter), body stiffness, stiffness of the floor (hard wood or carpeted), and pre-impact movements to arrest the fall. Not only the available models have limitations in considering all these parameters for simulating the fall and predicting the impact force, impact force itself is not appropriately considered as a risk factor in hip fracture risk assessment tools. This study reviewed the researches that investigated the range of impact force in a sideways fall and the range of femoral fracture force. Our review showed that the median value of the

femoral force in an unexpected fall from sideways for an average individual (5200 N) is well within or even beyond the range of the femoral fracture force measured in experimental studies (3500 N). Therefore, accurate prediction of fall-induced impact force is of great significance in developing hip fracture risk assessment tools, although it has not received as much attention as the bone strength did.

The mostly used formulas for predicting the impact force only consider body weight, body height, and hip soft tissues. However, there are many parameters that can be considered in modeling of human body falling to improve prediction of the impact force. This review divided these parameters to two categories: (1) the parameters that can be determined subject-specifically and (2) the parameters that may significantly differ from fall to fall for an individual and cannot be considered subject-specifically.

Body segment anthropometric parameters and body stiffness/damping properties are the effective factors that can be taken into account subject-specifically. Researches have been conducted to determine the anthropometric parameters from clinical images with an appropriate accuracy. Also, many studies have been devoted to investigate body stiffness/damping properties from experimentations. However, we conclude that accurate subject-specific determination of stiffness/damping properties is complex since they are affected by many parameters, including fall direction, condition of muscles (relaxed/contracted), landing surface and the portion of muscles and ligaments that contribute to the impact, abdominal and thoracic viscera, and flexural deformation of the spine. So far, trochanteric soft tissue thickness has been considered as a determinant of the attenuated impact force while other factors can also remarkably affect the impact scenario. This review concludes that more research is required on this topic. Future studies are in need to investigate how sex affects the attenuated force per each millimeter of hip soft tissues, how the portion of fat/muscle content affects the attenuated force, how to take the flexural deformation of spine into account, and how the landing surface and the portion of muscles that contribute to the impact can be considered in determining the impact force.

Pre-impact movements to arrest the fall as well as cause and initial conditions of fall are the parameters that cannot be considered subject-specifically as they may vary from fall to fall for an individual. These parameters significantly affect the kinematics of fall and subsequently the magnitude of the impact force. The lack of information about the complicated effect of cause/initial conditions of fall and pre-impact movements on the magnitude of the impact force is due, in part, to the limitations in performing real-life fall experiments because of the ethical and safety reasons. Therefore, this study suggested using collected database of real-life falls to extract these parameters. As initiated by Robinovitch et al. [74], video-captured falls, experienced by the elderly at long-term care facilities, can be used to determine the most frequent causes of falls. Recorded videos can also be analyzed to extract the common initial conditions of falls and the frequent pre-impact movements to arrest the fall. Results of analyses can then be implemented in human body modeling to more realistically simulate the human body falling and determine the impact force. This review also concludes that determination of joint torques is quite complicated and more research is required on this topic.

Not only there is a need for developing an accurate model for predicting the fall-induced impact force, impact force itself should be more appropriately considered in hip fracture risk assessment tools. FE models are the only tools that explicitly take the impact force into consideration. An improved model for determining the impact force can be a step forward to more appropriately consider the impact force in hip fracture risk assessment.

Due to the worldwide increasing rate of hip fractures, developing an accurate model for predicting the impact force and adopting the model in hip fracture risk assessment tools can be considered as a crucial priority for biomechanical researches. It could improve the accuracy of identifying patients at high risk of fracture who may benefit from in-time treatment.

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