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ORIGINAL ARTICLE

Estimation of effective mass of longish rigid instruments in head impacts

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Abstract Impacts to the head are a common form of body violence and thus a relevant legal medical issue. Biomechanical assessment of injury potential has been traditionally based on qualitative analysis and experience. The aim of this study was to collect benchmark data that would facilitate the assessment of the maximum force in head impacts with longish rigid instruments. Series of measurements were performed with a specially designed modifiable impactor, and the relationship between its inertial properties and its effective mass during the impact was studied. The effective mass was defined as the amount of point mass that would, if exposed to the same velocity change as the striking end of the instrument, produce the same area under the force-time curve as the impactor. The results show that the effective mass decreases from approximately 100% of the total body mass for very short impactors to about 50% for longer (approximately 70 cm) impactors. No influence of the hand/grip force on the effective mass of the impactor was found if it was used in a hammer-like manner; other striking techniques can lead to substantial increase of the effective mass attributable to the hand/grip force.

Keywords Head impact · Biomechanics · Effective mass · Injury potential

Introduction

According to the German police, more than 500,000 criminal assaults were registered in 2006 with 171 deaths. Moreover, the number of body violence cases showed an increasing tendency [1]. Analyses of criminal assaults indicated that the usage of blunt instruments for battery amounts to approximately 10% with blows most frequently (approximately 66%) launched against the head [2]. Similar incidence rate (9%) in the usage of blunt instruments was reported by [3]. Thus, blows to the head constitute an important forensic biomechanical issue.

In forensic biomechanical reconstruction, it is required to assess the injury potential of the assault and to evaluate legally relevant problems. Forensic biomechanical expertise of events resulting in injuries (accidents, assaults etc.) takes into account objective evidence as well as testimonies of the concerned persons (assailant, victim) and the witnesses (if available). Based on the analysis and assignment of individual traces on all concerned systems (victim, assailant, surrounding structures), the temporal and spatial aspects of the event are clarified and loads imposed to the human body are assessed in both qualitative and quantitative way (an example for this is bicycle collisions; this procedure is described in [4]). On the other hand, the objectively incurred injuries are assessed as to their mechanisms and magnitude of loading. By putting together the information about the course of actions and about the injuries suffered, the causality between the assault and the injuries can be assessed and other legally relevant questions answered.

Impact to the human head is a very complex phenomenon. Its detailed quantitative analysis would have to consider all the characteristics of the head which are individual and thus not exactly ascertainable as well as the mechanical properties of the impactor and the kinematics of its movement prior to impact. Even though highly sophisticated finite elements models are available, many input parameters are not known. Even comparably simple analytical solutions (i.e. differential equations governing impact of smooth rigid bodies) require

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the knowledge of many parameters that can only be roughly estimated. Thus, the use of complex laws of mechanics would, after a demanding computation, lead to a solution compromised by potential errors with respect to many input parameters. For practical use in forensic biomechanics, a very robust and simple method of contact force estimation is based on Newton's Second law applied over time. The area under the force-time curve is assumed to be equal to the change of the linear momentum, i.e. to the product of mass and change of velocity of the impacting object

$$\int_{t_1}^{t_2} F(t)\mathrm{d}t = m\Delta v,$$

where $F(t)$ denotes the contact force, m mass, Δv change of velocity of the impacting object and t_1 and t_2 designate the points of time when the impact begins and ends, i.e. (t_2–t_1) is the impact duration. The maximum acting contact force can be closely approximated [5] by assuming a triangular shape of the force-time curve yielding

$$\mathrm{F}_{\max} \approx 2m\Delta v/\Delta t,$$

where $F_{\max}$ denotes the maximum contact force and Δt the contact duration.

The contact duration depends on the material properties of both the head and the impacting object. Head impacts with rigid objects last several milliseconds (3–7 ms for head impact on hardwood steps [5]); impacts with deformable objects last generally longer.

Because of contact elasticity, the change of velocity of the impactor can reach higher values than the impact velocity. In inverted drop tests as well as pendulum impacts (some of the tests slightly padded; [6]) it was found that the coefficient of restitution (ratio of velocities after and before impact) for head impacts reaches values around 0.22. Thus, the Δv of the impactor can be as high as 1.22 times the impact velocity (if the head is supported or the mass of the impactor is significantly lower than the head mass). The impact velocity is estimated from the available testimonies and evidence or its whole range of possible values is considered.

In central impacts of rigid objects (i.e. the line of contact force goes through the centre of gravity (COG) of the object) the change of momentum is described by the mass of the whole object times the velocity of its COG. However, in assaults the striking instruments are mostly longish (bats, rods, bottles etc.) and the impact occurs at their end roughly perpendicular to the long axis of the instrument. Thus, the impact configuration is highly eccentric.

In order to be able to use the velocity v of the striking end of the instrument for the computation of the maximum impact force $F_{\max}$, the concept of the effective mass m_{eff} must be introduced. It designates the amount of point mass that would, if exposed to a velocity change of Δv, produce the same area under the force-time curve as the striking instrument. The above equation can thus be reformulated as

$$\mathrm{F}_{\max} \approx 2\mathrm{m}_{\mathrm{eff}}\Delta v/\Delta t$$

In case of a freely moving impacting object, the effective mass is a fraction of its total mass (equal to it in central impacts of rigid bodies and potentially higher if there are constraint forces limiting the free movement of the impactor such as grip forces). In assaults, the movements of the striking instrument are constrained by the arm/hand of the assailant and the question arises in what way and to what extent the force transmission is altered—by the mass of the upper extremity swinging along with the instrument and/or by grip force restraining its movements during and after impact. The aim of the study presented in this paper was to assess the influence of the hand/arm on impact characteristics and to establish benchmark data regarding the effective mass of longish rigid striking instruments.

Problems of similar nature were studied in sports biomechanics. The grip tightness and its influence on post-impact ball velocity was analysed in tennis and baseball. It was found that grip firmness affects the post-impact ball velocity when using aluminium bats; the values were reportedly enhanced by a tight grip. Interestingly, grip tension had no significant effect when using wooden bats [7]. Several investigations of tennis strokes suggested that the level of grip tension affects the rebound velocity and the impulse imparted to a ball [8,9]. However, the influence of grip firmness on ball rebound was questioned by other findings suggesting that the coefficient of restitution of the racket and the ball were independent of the level of grip pressure [10].

Regarding heading in football, [11] found out that the effective mass of the head cannot be significantly influenced by contraction of the neck muscles.

Methods

Laboratory tests were performed simulating an impact of a bar-shaped object onto a human head. A special device was used to represent the elasticity of the skull. An elastic (Tecamid®) rod was attached to an aluminium platform by movable supports; the distance between the supports, and thus, the stiffness of the rod was adjusted to correspond with the human skull as found in experiments. The average rod stiffness found in the tests was 1,754 N/mm, which is well in agreement with the data for low-area impacts [12]. The device illustrated in Fig. 1 was placed on a Kistler force plate, and the reaction force during the impact was sampled with a frequency of 5,000 Hz. The impulse

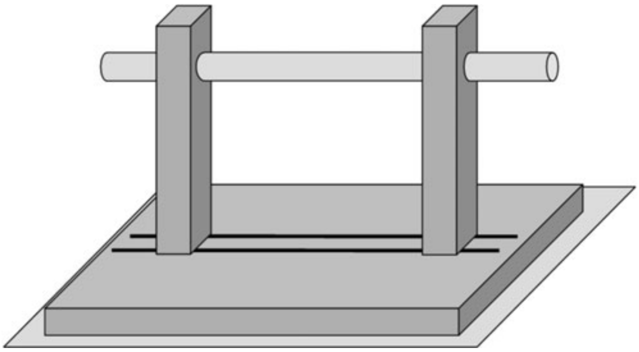


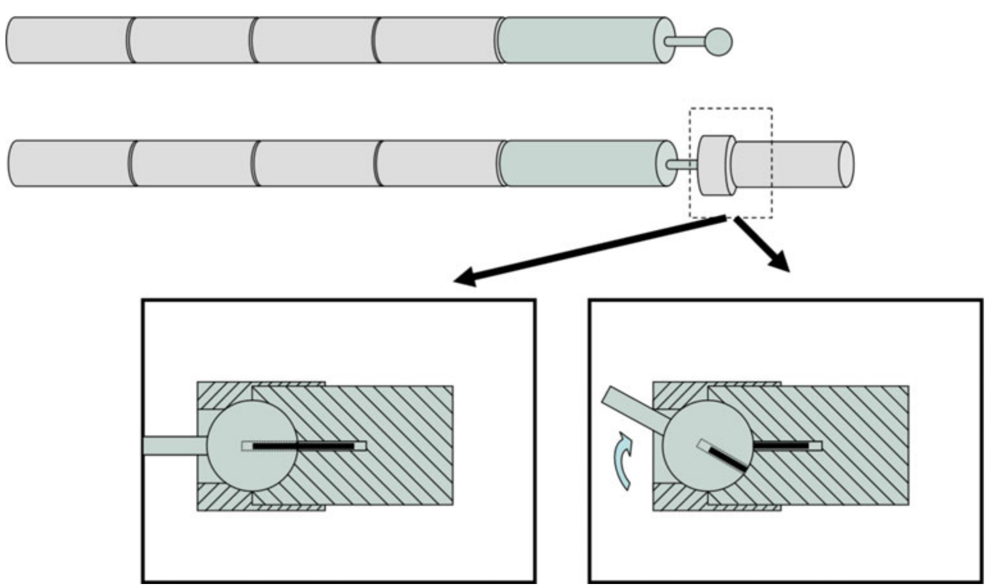
Fig. 1 The device for head impact simulation. The stiffness of the rod can be tuned by adjusting the distance between the supports

delivered by the impactor was computed by integrating the force over time signal.

A specially designed bar-shaped striking instrument (further referred to as impactor) was used in the study. The main body (the striking part) consisted of two to five pieces of hollow aluminium cylinders with the diameter of 3 cm that could be screwed together. The basis cylinder—the handhold—had a length of 12.9 cm and a mass of 264 g, the rest were each 10 cm long with the mass of 102 g. This way the length of the impactor was adjustable in 10 cm steps and the relationship between its geometrical and inertia properties and effective mass could be established. At the end of the handhold, there was a ball joint and an additional piece could be attached to the main body (see Fig. 2). When held at the additional piece, the main body could rotate around the centre of the joint and thus the impact was not influenced by the hand/arm of the volunteer. During the swing phase, the neutral position in this joint (i.e. the longitudinal axes of all segments in a line) was secured by a thin plastic bolt that was sheared off immediately at the impact and the impactor was then allowed to rotate freely in the ball joint. Two bolt types were used—a thicker one (diameter, 4 mm) for higher mass/length of the impactor and a thinner one (diameter, 3 mm) for low mass/moment of inertia. Two configurations were compared in order to assess the influence of the upper extremity on the impact characteristics. In the first one, the impactor was held at its handheld and the impact was influenced by the hand/arm of the volunteer. In the second one, the impactor was held at the additional piece with the striking part allowed to rotate and thus free of inertial and/or grip influence of the hand/arm of the volunteer. Please note that the weight of the mass as well as the length of the striking part was always the same but the overall length of the instrument varied.

A single male subject (42 years, body height 173 cm, body mass 73 kg) volunteered in this study. His task was to hit the elastic rod of the device imitating the properties of human skull at its middle part with the end piece of the impactor. The movement was similar to the use of a hammer (i.e. the blow was performed downwards in the sagittal plane, see Fig. 3). After each trial, the volunteer was informed about the maximum reaction force reached and was asked to use only as much effort as not to exceed a 6 kN border. The reason for this was that higher forces would be out of the region of elastic skull deformation, fractures would become very likely [13,14]. Care was taken that the impactor hit the plastic rod at the height of the point, i.e. approximately 5 cm from its end. If the impact was located more than 1 cm from this point, the measurement was repeated. There were no other constraints regarding the striking movement. When the bolt sheared off during the swing, the measurements were repeated and the volunteer asked to perform the movement in the same way that is natural to him.

Fig. 2 The impactor. The additional piece is connected by a ball joint to the impactor in some configurations. It is kept aligned by a thin bolt during the backswing; at contact, the bolt is sheared off immediately and the impactor body can rotate freely in the joint



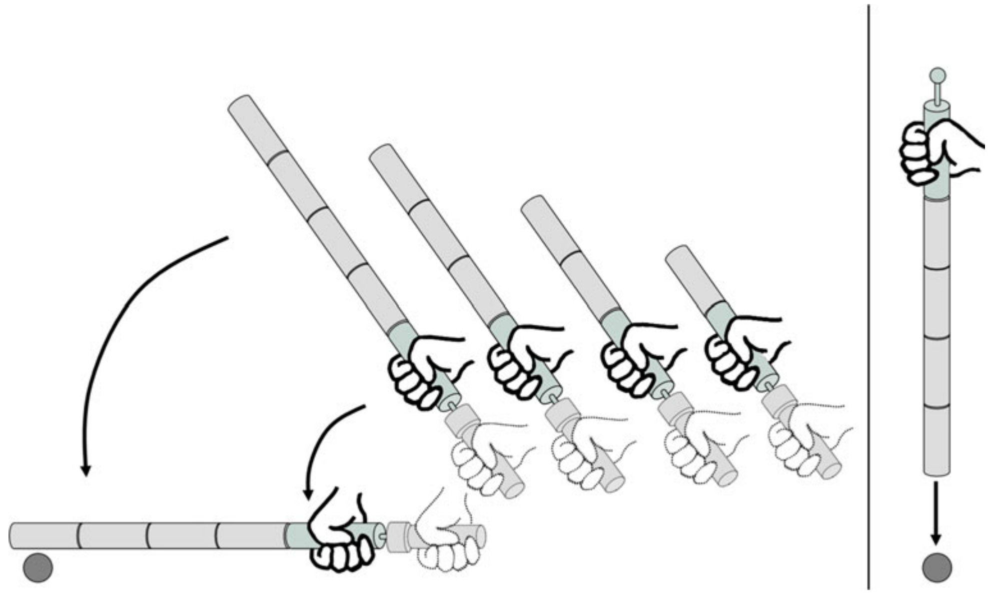


Fig. 3 Swing techniques used in the study. On the *left*, the hammer-like movement, i.e. the impactor is used like a hammer and hits the rod in a direction perpendicular to the shaft. The impactor body of variable length is held either at its end or on the additional piece connected by a ball joint. On the *right*, the stab-like technique is depicted, i.e. the impactor moves along its longitudinal axis

The movement of the striking end of the instrument was recorded using a high-speed video system (MotionBLITZ® Cube123PL, Nikon AF NIKKOR camera) with a sampling frequency of 2,000 Hz. A reflective marker was placed at 5 cm from the end of the impactor and its position was tracked. From these data, the velocity before and after impact was determined as the mean velocity over the last five frames (0.01 s) prior to contact and the first five frames after contact.

Since many striking instruments used in assaults are not homogeneous, i.e. the mass is not distributed evenly and the centre of gravity is moved towards one end (most of the times to the striking end like in bats, hammers etc.), further measurements were performed with additional mass attached at the striking end of the instrument. Two additional masses were used—136 g and 744 g representing a slight (a baseball bat etc.) and a strong (hammer etc.) shift of the COG towards the striking end. The centre of gravity of the additional mass was in both cases 5 cm from the end of the impactor and thus over the impact point. Finally, another kind of swing motion was considered—the impactor was held vertically and the movement was performed in a stab-like manner, i.e. in the longitudinal axis of the impactor downwards (see Fig. 3). In each configuration, two trials were performed.

Results

The results are summarised in Tables 1, 2, 3 and 4. The measured values are the velocities of the striking end of the impactor immediately before and after contact with the elastic rod and the reaction force maximum. The impact duration was derived from the force-time curve. The coefficient of restitution is the ratio between the rebound and impact velocity, the impulse was computed by integrating the force-time signal during contact. The effective mass was computed as $m_{\mathrm{eff}} \approx F_{\max}\Delta t/2\Delta v$, where $F_{\max}$ is the force maximum during the impact, Δt is contact duration and Δv is the difference between the impact and rebound velocity (the latter is negative, i.e. Δv is the sum of the two absolute values). The effective mass was expressed both as an absolute value in kilograms and as a percentage of total impactor mass.

It must be noted that in some trials impactor oscillations were observed, and thus rebound velocity measurement was impaired. For this reason, some configurations with higher impactor lengths are missing. The measured maximum force data suggest that all the trials stayed within the range of elastic skull deformations.

Figure 4 shows the relationship between the mass distribution of the impactor represented by the distance between the centre of the grip area (point of grip (POG), 5 cm from the end of the impactor) and the COG and its effective mass (average of two values for each configuration) at impact. Though the data show some scatter, there is a trend toward lower effective mass with increasing distance between the COG and the POG. The effective mass decreases from approximately 100% of the total body mass for very short impactors to about 50% for longer impactors. Please note that the distance between the grip area and the centre of mass of 30 cm corresponds to impactor length of 70 cm (for a homogeneous body).

The (lack of) influence of the hand/grip force on the effective mass can also be assessed from Fig. 4. In all cases the effective mass was slightly higher in trials with the impactor held at the additional piece connected to the base with the ball joint. However, the differences were small.

Table 1 The impact parameters—impactor without additional weight, hammer-like movement

Trial	Joint	l_{imp} (m)	m_{imp} (kg)	F_{max} (N)	Imp. dur (s)	v_{imp} (m/s)	v_{reb} (m/s)	k	Impulse (Ns)	m_{eff} (%)	Aver. m_{eff} (%)
1	No	0.229	0.366	1,743	0.0024	4.6	4.0	0.87	2.18	70	69.5
2	No	0.229	0.366	1,857	0.0026	5.1	3.9	0.76	2.27	69	
3	No	0.329	0.468	2,513	0.0026	5.7	4.5	0.79	3.05	64	64.5
4	No	0.329	0.468	2,463	0.0024	6.0	4.2	0.70	3.11	65	
5	No	0.429	0.570	2,511	0.0028	5.8	4.4	0.76	3.25	56	60.5
6	No	0.429	0.570	3,092	0.0028	6.3	4.8	0.76	4.13	65	
7	No	0.529	0.672	2,951	0.0030	4.9	4.4	0.90	3.89	62	61.0
8	No	0.529	0.672	2,514	0.0030	5.0	3.5	0.70	3.44	60	
9	No	0.629	0.774	3,324	0.0028	6.5	5.0	0.77	4.31	48	47.5
10	No	0.629	0.774	2,955	0.0030	6.2	4.7	0.76	3.95	47	
11	Yes	0.429	0.570	2,313	0.0030	5.0	3.9	0.78	3.06	60	61.0
12	Yes	0.429	0.570	3,252	0.0028	6.3	4.9	0.78	3.97	62	
13	Yes	0.329	0.468	2,217	0.0026	5.3	3.9	0.74	2.97	69	75.5
14	Yes	0.329	0.468	3,262	0.0032	6.0	5.0	0.83	4.21	82	
15	Yes	0.229	0.366	2,487	0.0026	6.5	4.8	0.74	3.23	78	81.5
16	Yes	0.229	0.366	2,249	0.0024	5.4	4.1	0.76	2.94	85	

No the impactor was held on its end, *yes* the impactor was held on the additional piece connected by a ball joint, l_{imp} length of the impactor, m_{imp} mass of the impactor, F_{max} maximum of the contact force, *Imp. dur.* impact duration as derived from the force signal, v_{imp} impact velocity, v_{erb} rebounce velocity, *k* coefficient of restitution, m_{eff} effective mass of the impactor, *Aver.* m_{eff} average value of the effective mass

In the stab-like impacts, the effective mass of the impactor was apparently not affected by its length (see Fig. 5).

Discussion

The aim of our study was to establish benchmark data of the effective mass of a rigid striking instrument for forensic biomechanical analysis of head impact. Our results indicate that the effective mass of the striking instrument depends on the location of the COG with respect to the POG. Consequently, the load potentially imposed to the head can be estimated for various striking instruments.

In real-world forensic analyses, the individual biomechanical properties of the persons involved are not known and reference values established through experiments and reported in the literature vary considerably. In addition, the impact velocity depends on many individual and situational

Table 2 The impact parameters—impactor with an additional weight of 136 g, hammer-like movement

Trial	Joint	l_{imp} (m)	m_{imp} (kg)	F_{max} (N)	Imp. dur (s)	v_{imp} (m/s)	v_{reb} (m/s)	k	Impulse (Ns)	m_{eff} (%)	Aver. m_{eff} (%)
1	No	0.229	0.502	2224	0.0026	4.5	2.8	0.62	3.04	83	86.5
2	No	0.229	0.502	2,396	0.0026	4.5	2.7	0.60	3.25	90	
3	No	0.329	0.604	3,077	0.0030	5.1	3.6	0.71	4.26	81	78.0
4	No	0.329	0.604	2,598	0.0028	4.5	3.2	0.71	3.48	75	
5	No	0.429	0.706	2,971	0.0028	4.8	3.5	0.73	4.01	68	71.0
6	No	0.429	0.706	2,904	0.0028	4.3	3.1	0.72	3.87	74	
7	No	0.529	0.808	2,988	0.0030	4.8	3.2	0.67	4.25	66	67.0
8	No	0.529	0.808	3,373	0.0032	5.1	3.6	0.71	4.81	68	
9	No	0.629	0.910	3,249	0.0032	4.9	3.1	0.63	4.75	65	64.0
10	No	0.629	0.910	3,164	0.0032	4.6	3.3	0.72	4.54	63	
11	Yes	0.429	0.706	4,057	0.0030	6.0	4.5	0.75	5.89	79	78.5
12	Yes	0.429	0.706	4,186	0.0032	6.2	4.8	0.77	6.02	78	
13	Yes	0.329	0.604	4,125	0.0030	5.8	4.3	0.74	5.63	92	92.0
14	Yes	0.329	0.604	3,001	0.0028	4.6	3.0	0.65	4.24	92	
15	Yes	0.229	0.502	3,471	0.0032	5.1	3.7	0.73	4.72	107	110.0
16	Yes	0.229	0.502	4,170	0.0032	6.2	4.8	0.77	6.23	113	

No the impactor was held on its end, *yes* the impactor was held on the additional piece connected by a ball joint, l_{imp} length of the impactor, m_{imp} mass of the impactor, F_{max} maximum of the contact force, *Imp. dur.* impact duration as derived from the force signal, v_{imp} impact velocity, v_{erb} rebounce velocity, *k* coefficient of restitution, m_{eff} effective mass of the impactor, *Aver.* m_{eff} average value of the effective mass

Table 3 The impact parameters—impactor with an additional weight of 754 g, hammer-like movement

Trial	Joint	l_{imp} (m)	m_{imp} (kg)	F_{max} (N)	Imp. dur (s)	v_{imp} (m/s)	v_{reb} (m/s)	k	Impulse (Ns)	m_{eff} (%)	Aver. m_{eff} (%)
1	No	0.229	1.120	2,194	0.0044	2.3	1.6	0.7	4.33	99	102.0
2	No	0.229	1.120	4,314	0.0048	4.0	3.0	0.75	8.23	105	
3	No	0.329	1.222	4,452	0.0050	3.7	2.9	0.78	8.64	107	103.5
4	No	0.329	1.222	4,255	0.0040	3.7	2.7	0.73	7.86	100	
5	No	0.429	1.324	4,268	0.0038	3.7	3.2	0.86	7.59	83	86.5
6	No	0.429	1.324	4,709	0.0040	3.8	3.2	0.84	8.30	90	
7	No	0.529	1.426	5,759	0.0040	4.5	3.8	0.84	10.71	90	88.0
8	No	0.529	1.426	4,692	0.0036	3.7	3.1	0.84	8.33	86	
9	No	0.629	1.528	5,528	0.0050	4.2	3.6	0.86	11.16	94	89.0
10	No	0.629	1.528	4,892	0.0040	3.9	3.3	0.76	9.21	84	
11	Yes	0.429	1.324	4,621	0.0038	3.7	3	0.78	8.78	99	102.5
12	Yes	0.429	1.324	5,012	0.0048	4.0	3.2	0.78	10.07	106	
13	Yes	0.329	1.222	5,195	0.0050	4.3	3.4	0.79	10.51	112	111.5
14	Yes	0.329	1.222	5,324	0.0050	4.4	3.6	0.82	10.89	111	
15	Yes	0.229	1.120	5,677	0.0040	4.8	3.9	0.81	10.82	111	107.0
16	Yes	0.229	1.120	5,427	0.0036	4.7	3.9	0.83	9.96	103	

No the impactor was held on its end, *yes* the impactor was held on the additional piece connected by a ball joint, l_{imp} length of the impactor, m_{imp} mass of the impactor, F_{max} maximum of the contact force, *Imp. dur.* impact duration as derived from the force signal, v_{imp} impact velocity, v_{erb} rebounce velocity, *k* coefficient of restitution, m_{eff} effective mass of the impactor, *Aver.* m_{eff} average value of the effective mass

factors and can never be regarded undisputable. From this point of view, the estimation of the effective mass seems to be reasonably accurate and can be recommended for practical use. Calculating the regression between the effective mass (expressed as a percentage of the total impactor mass) and the distance between COG and POG (in cm) results in $y = -1.63 \times +100.8$ with a coefficient of determination equalling 0.72 (Fig. 5).

An important finding is that the effective mass of the impactor, and thus, the imposed load in terms of maximum contact force is not significantly influenced by the grip force and/or by the mass of the hand/arm if the impactor is swung in the hammer-like manner. This can be attributed to the short impact duration and the limited motion of the impactor during contact. In the study presented, only contact forces up to the approximate biomechanical tolerance of the skull were produced. In general, with increasing impact severity the influence of the grip force should decrease even further. However, high contact force would result in skull fracture, and thus, the contact characteristics would be altered due to plastic deformation.

The regression line represented by the above-mentioned equation suggests that as the distance between COG and POG goes to its natural lower limit (i.e. zero, the centre of gravity would coincide with the location of impact), the effective mass of the impactor approaches 100%. This is in accordance with the fact that the hand does not influence the effective mass significantly. Approaching its other limit, i.e. for large distances between COG and POG, the equation would predict negative effective masses. This does not reflect the physical laws—one would expect continuous decrease of the relative effective mass but it

Table 4 The impact parameters—stab-like movement

Trial	Joint	l_{imp} (m)	m_{imp} (kg)	F_{max} (N)	Imp. dur (s)	v_{imp} (m/s)	v_{reb} (m/s)	k	Impulse (Ns)	m_{eff} (%)	Aver. m_{eff} (%)
1	No	0.229	0.366	1,909	0.0032	3.2	0.0 (2.4)	0 (0.75)	3.14	153	156.5
2	No	0.229	0.366	2,641	0.0036	4.0	0.0 (3.0)	0 (0.75)	4.10	160	
3	No	0.329	0.468	3,134	0.0034	4.1	0.0 (3.1)	0 (0.75)	5.21	155	159.5
4	No	0.329	0.468	2,858	0.0042	3.9	0.0 (2.9)	0 (0.75)	5.23	164	
5	No	0.429	0.570	3,683	0.0038	3.8	0.0 (3.1)	0 (0.75)	6.48	170	166.0
6	No	0.429	0.570	3,373	0.0038	3.8	0.0 (3.1)	0 (0.75)	6.19	162	
7	No	0.529	0.672	4,458	0.0036	4.0	0.0 (3.0)	0 (0.75)	7.40	157	157.0
8	No	0.529	0.672	3,471	0.0036	3.3	0.0 (2.5)	0 (0.75)	6.13	157	

No the impactor was held on its end, *yes* the impactor was held on the additional piece connected by a ball joint, l_{imp} length of the impactor, m_{imp} mass of the impactor, F_{max} maximum of the contact force, *Imp. dur.* impact duration as derived from the force signal, v_{imp} impact velocity, v_{erb} rebounce velocity, *k* coefficient of restitution, m_{eff} effective mass of the impactor, *Aver.* m_{eff} average value of the effective mass

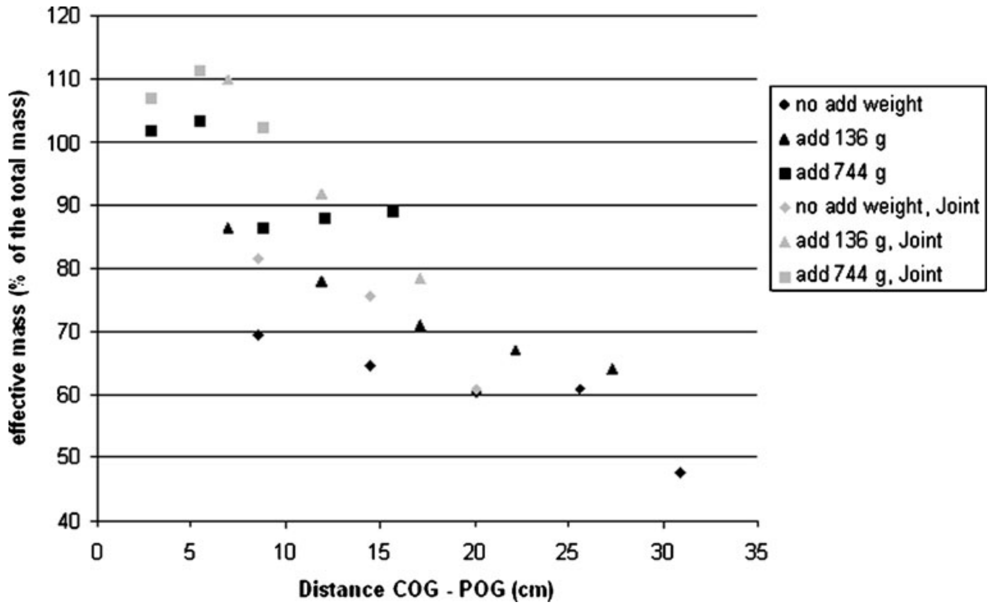


Fig. 4 The relationship between effective mass of the impactor and its inertial properties represented by the distance between the centre of gravity and point of grip. In addition, the respective values obtained in configurations with the impactor held on the additional piece are included (coloured *grey*) no added weight—basic form of the impactor, no additional weight used; add 136 g—additional weight of 136 g was attached at the striking end of the impactor add 744 g—additional weight of 744 g was attached at the striking end of the impactor

would only approach zero as the COG–POG distance would grow beyond all limits. The above presented relationship is only meaningful for COG–POG distances within the range of this study and only for (approximately) rigid objects. In case of longer or non-rigid impactors, other physical phenomena (such as oscillations and plastic deformations) would have to be considered that would exert significant influence on contact characteristics and thus on the effective mass.

Figure 5 suggests that other than linear relationships between the effective mass and its mass distribution might exist. However, taking into account the already mentioned uncertainty involved in forensic reconstructions, the linear model seems to be sufficient and the most suitable for practical use.

The predefined impact location 5 cm from the free end turned out to be the one spontaneously chosen by the volunteer for all considered impactor lengths, and thus, it seems to be the correct assumption for real-world situations. An exception hereof might represent objects with special shapes that lead the assailant to use it otherwise; for example, a relatively thin baton with a robust knob at its end or a hammer would automatically be swung such that the head would be hit by the very end of the instrument.

As opposed to the majority of the real-world situations, in the experimental setup the impacted object did not move. Depending on the impact severity and direction, the head (with a mass of approximately 5 kg in adult men and slightly less in women) is accelerated in the direction of the

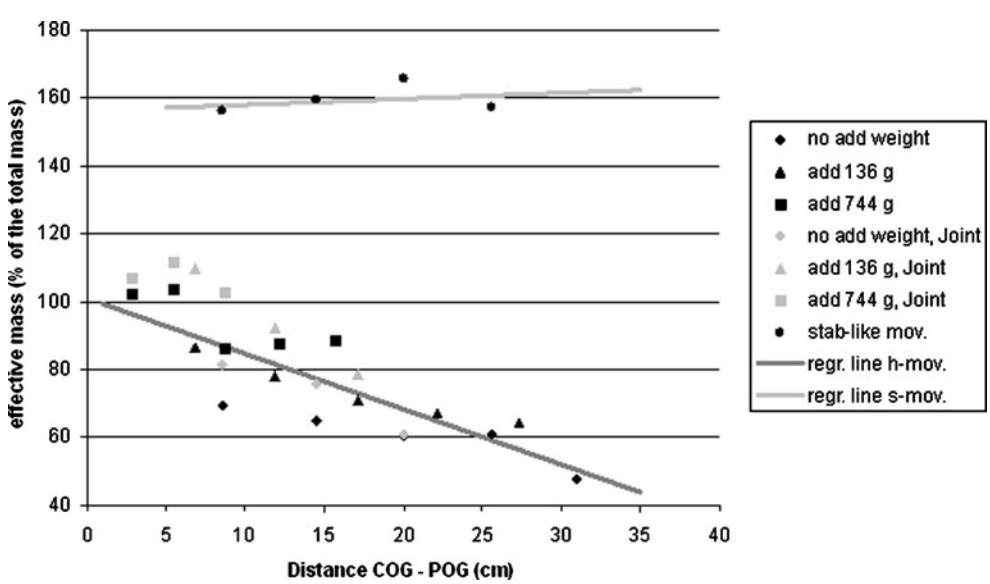


Fig. 5 The effective mass of the impactor in both hammer- and stab-like techniques and the respective regression lines

contact force. However, this does not influence the effective mass of the impactor.

Our experimental setting did not account for the skin. Its presence would presumably slightly influence the contact properties—the contact duration would be slightly longer and the coefficient of restitution lower due to skin compression and the associated damping effects.

The force plate registered the force in all three directions and the resultant was computed, i.e. potential deviations from the intended vertical blow direction could not bias the output.

As expected, the distance between the COG and POG did not influence the effective mass in impacts after stab-like movement. However, in this constellation the coupling between the arm and the impactor alters the mechanical parameters of the impact and the subsequent movement of the impactor substantially—no rebound was observed and the effective mass of the impactor reached up to 300% of its total mass.

The contact duration derived from the force data and shown in Table 4 was in the same range as in hammer-like swing motion in spite of the fact that no observable rebound was detected. The force data show that the grip force influences the interaction not to the extent suggested by the missing rebound; significant contact force occurred only for several milliseconds. It leads to the assumption that if the impactor had been released at the point of maximum contact force, the rebound would have taken place; in other words, the motion of the impactor after the contact was restricted by the hand/arm and the observed zero rebound velocity does not reflect the effective mass at impact. In order to obtain a more realistic estimate of the effective mass, it can be assumed that the rebound took place and was characterised by the same coefficient of restitution as found in the hammer-like motion measurements; this assumption seems reasonable since both contact partners are the same. Assuming an average k of 0.75, the effective mass of the impactor reaches approximately 160% of its total mass (see Table 4 and Fig. 5; the regression line in Fig. 5 is expressed by the equation $y = 0.147 \times +157.2$, the coefficient of determination is as low as 0.06).

Surprisingly, the effective mass expressed as a percentage of the actual impactor mass did not seem to be affected by its increase associated with growing length (the actual mass varied between 0.366 and 0.672 kg and absolute values of the effective mass between 0.56 and 1.06 kg) in stab-like blows. Based on the physical laws, one would expect that with increasing actual impactor mass (in case of the used instrument, the actual mass grew with increasing COG–POG distance) its effective mass would decrease (asymptomatically toward 100% for huge masses), i.e. the influence of the grip force would decrease. The fact that we did not observe such trend could be attributed to increased grip force associated with the handling of heavier instrument; also, the range of impactor lengths/masses was probably not broad enough for this trend to show.

The results show that though the stab-like movement would generally result in lower impact velocity, due to higher effective mass the maximum contact force can reach similar or even higher values. As a result, from the forensic biomechanical point of view this kind of attack cannot be regarded as less dangerous even though it is typically associated with lower impact velocity.

Conclusions

In hammer-like blows with a longish rigid instrument, effective mass (defined as the amount of point mass that would, if exposed to a velocity change of Δv, produce the same area under the force-time curve as the striking instrument) decreases with increasing impactor length; if the distance between the centre of gravity of the impactor and the point of grip on the instrument reaches approximately 0.3 m (corresponding to a length of 70 cm of a homogenous impactor) the effective mass accounts for approximately 50% of the impactor mass.

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References

1. Kriminalstatistik (2009). Available at: http://bka.de/pks. Accessed 15 March 2010
2. Missliwetz J (1990) Circumstances and injury pattern of intentional bodily injuries (especially weapon injuries). Beitr Gerichtl Med 48:299–307
3. Eppendorf Kirstin (2000) Gesichtsschädelverletzungen durch Rohheitsdelikte - Häufigkeit, Ursachen, soziale Begleitumstände und ökonomische Belastungen. Dissertation, Martin Luther University of Halle-Wittenberg, Hale, Germany
4. Graw M, König HG (2002) Fatal pedestrian-bicycle collisions. Forensic Sci Int 126(3):241–247
5. Goldsmith W, Plunkett J (2004) A Biomechanical Analysis of the Causes of Traumatic Brain Injury in Infants and Children. Am J Forensic Med Pathol 25:89–100
6. Viano DC, Parenteau ChS (2008) Analysis of head impacts causing neck compression injury. Traffic Inj Prev 9:144–152
7. Weyrich AS, Messier SP, Ruhmann BS, Berry MJ (1989) Effects of bat composition, grip firmness, and impact location on postimpact ball velocity. Med Sci Sports Exerc 21(2):199–205
8. Elliot BC (1982) Tennis: the influence of grip tightness on reaction impulse and rebound velocity. Med Sci Sports Exerc 15 (5):348–352
9. Hatze H (1976) Forces and duration of impact, and grip tightness during the tennis stroke. Med Sci Sports 8(2):88–95
10. Watanabe T, Ikegami Y, Miyashita M (1979) Tennis: the effects of grip firmness on ball velocity after impact. Med Sci Sports 11 (4):259–361

11. Schneider K (1985) Der Einfluß motorischer und mechanischer Stoßbedingungen auf das Verletzungsrisiko beim Fußball-Kopfstoß. Sportwissenschaft 15:183–192
12. Allsop DL, Perl TR, Warner ChY (1991) Force/deflection and fracture characteristics of the temporo-parietal region of the human head. In: Proceedings 35th Stapp Car Crash Conference. pp 199–218
13. Nahum AM, Gatts JD, Gadd CW, Danforth JP (1968) Impact tolerance of the face and skull. Proceedings 12th Stapp Car Crash Conference: 302–317
14. Advani S, Powell W, Huston J, Ojala S (1975) Human head impact response - experimental data and analytical simulations. In: Proceedings International IRCOBI Conference on the Biomechanics of Impact. pp 153–162