A Robotics Perspective on Experimental Injury Biomechanics of Human Body Upper Extremities

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***Abstract*— Achieving a better understanding of injury mechanisms and significant factors affecting safety in robotics requires large amounts of ground-truth, medically classified biomechanical information. Representing impact data-sets as relationships between input parameters (reflected mass, velocity, injury) for different impact curvatures and test setups versus injury was previously proven to be very efficient for ensuring human safety through real-time robot control algorithms. However, such a data-driven safety framework requires enough data-sets to cover all practical collision incidents. In this work, resulting data-sets from our extensive literature review process on the human body upper extremities are summarized. We also adapt our previously developed robotics-compatible representations in terms of standardized testing setups, impact scenarios, and collision cases to distinctly categorize impact-testing experiments on the human body upper extremities. Each category can then be differentiated by a minimal representative set of test arrangements to unify the description, storage, reporting, and comparison of experimental settings and results. This way, every new impact test/incident can be easily categorized and compared against others in the developed digital database.**

I. Introduction

In order to successfully introduce robots in everyday environments, dependability and human safety are primary concerns [1]. Human safety in robotic terms can be classified into two categories: psychological and physical safety. The first deals with ensuring that human interaction with robot does not cause excessive discomfort and stress for relatively long time periods [2]. The latter, physical safety, aims at preventing all unwanted human-robot contacts. Alternatively, if contact is an integral part of the task or it is inevitable for another reason, a safe physical human-robot interaction (pHRI) scheme must ensure that the forces exerted on human fall below limits of causing pain or injury [3]. To this end, the investigation of injury mechanisms and the development of safe mechanical designs, collision sensing and identification in addition to control strategies are ongoing topics in robotics research and many efforts have been taken until now e. g. [4]–[8]. Nonetheless, from the human body perspective, the possible injuries are hardly investigated. Two early studies were dealing with head and chest blunt impacts [9]–[11] employing dummies and human volunteers. Soft tissue injuries were examined for developing the collision detection methods in [12], [13], where a rather extreme case with a sharp knife impacting a pig cadaver specimen is demonstrated. Similar test subject was also used with other impact geometries [14]. In that work, the author proposed safety curves summarizing the impact results with parts of

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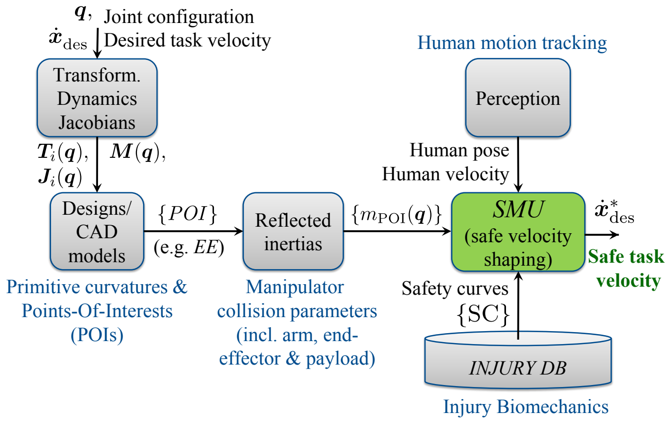


Fig. 1: The Safe Motion Unit (SMU) pipeline for ensuring biomechanical human safety of robot manipulators during pHRI [14].

porcine cadavers. Additionally, their experimental findings were used as informative safety data in the ISO/TR 23482-1 robotics standard [15], where also animal experimental data was introduced for other extremities. For example, some of these data were generated from impacting e. g. bear feet, and the standard used them to calculate the fracture tolerance of human foot bone. In general, benefits of animal impact testing in evaluating potentially injurious events within a crash environment involving a living human are still obvious, since the latter are best represented in a laboratory setting by a living organism [16]. Additionally, impact testing of living and dead animals has historically provided needed insight into controlling for differences between living humans and cadavers when developing injury response [17], [18].

Real human arm injury in robotics is only investigated lately in [19]. In that work mostly a dummy device is used and the resulting findings were compared against data from a single human volunteer experiment. Therefore, only very low injury severity levels were examined due to the obvious limiting safety regulations and ethical restrictions for human volunteer testing [18]. For this last reason and others[1], available injury biomechanics data on human body upper extremities in general, and more specifically for robotics applications, is very few and not well-classified nor ready for easy incorporation into robot controllers in safe pHRI. Furthermore, a real-time, safe robot control framework is needed for enhanced interactive tasks with the human co-worker. A unified safety framework that relies on biomechanics information to ensure human safety in case of collision is called the safe motion unit (SMU) [14], see Fig. 1.

[1]For example, in the automotive safety research injuries to arms and hands are not the top priority (like head and chest) [20], [21], and also due to the different use case there from robot manipulators.

The SMU limits robot velocity to a safe level based on robot dynamic properties (i.e. reflected inertia, geometry of potential collision points and velocity) and human injury database. Later, this algorithm was developed to a framework of global safety assessment called Safety Map [22] and the influence of a payload on maximum operational velocity and inertial parameters was investigated [23].

A. Problem definition and research contribution

Dexterous and complex dynamic manipulation in interactive tasks requires an accurate interplay of the involved human arm and robot (end-effector). This necessitates a tighter interaction space, where only safe motion trajectories and torques have to be exercised by the robot not to harm the human co-worker arm/hand. To fulfill this essential requirement without compromising the task productivity unnecessarily, precise human injury tolerance levels in terms of maximum force/torque or (angular) velocity/acceleration must be known. However, even with all generic and task-specific biomechanics data-sets experimentally generated for the human arm and hand injuries, there are many data-gaps and classified information about impacts on human upper extremities robotics perspective is still missing. Hence, our research tries to address this problem through the following contributions:

- An extensive literature review, classification, and digitization process has been completed for all the published literature on injury biomechanics of human upper extremities.
- Synopsis plots summarizing force and energy content in terms of mass versus velocity are introduced.
- All available descriptions of fracture/breaking and bending moments for upper extremities are collected and digitized.

II. The Safe Motion Unit

The SMU is composed essentially of three parts: (1) Perception of the human; (2) Injury biomechanics database; (3) Velocity-shaping task control. The perception module tracks human motion, detects each human body part (key-points), and measures its relative distance to the robot point-of-interest POI (e. g. end-effector EE). Consequently, the position of the human key-points and robot POI in the 3D Cartesian space, expressed in the same reference frame (e. g. at the robot base), have to be continuously monitored. A displacement vector capturing the relative distance can be constructed, and the relative velocity can also be estimated. The injury/safety information from the injury biomechanics database specifies the upper limits, by e. g. a suitable safety curve (per impactor curvature), under which the instantaneous robot speed should be kept. A straightforward variant of such curves encoding the injury/safety information for each curvature and impact settings is the simple linear regression. For such a linear curve, the upper limit on relative speed can be calculated from

$$v_{\max}(m) = \mathrm{reg\,.\,lim}\left[c_1\left(i,\mathbf{p}_i\right)m + c_2\left(i,\mathbf{p}_i\right), v_1, v_2\right], \qquad (1)$$

with $c_1(i,\boldsymbol{p}_i) < 0$ and $c_2(i,\boldsymbol{p}_i) > 0$ being the coefficients of the delimiting safety/injury curves for contact surface primitive i with its associated descriptive parameters $\boldsymbol{p}_i$, and v_1, v_2 denoting the cut-off minimum and maximum velocity. Respecting those safety limits, e. g. through a safe velocity controller as done in the SMU, guarantees a human-safe physical interaction, even during contact phases in case entirely unforeseen collisions occur. The complete algorithmic details of the SMU can be found in the seminal SMU paper by Haddadin et al. [14]. The SMU algorithm was recently extended into an offline framework for global safety assessment and planning called the Safety Map [22], and the influence of a payload on maximum operational velocity inertial parameters was also investigated lately [23].

III. Proposed Methodology

We aim to represent all the gathered injury biomechanics data from our extensive literature review process in a unified way such that a comparative, comprehensive and easily accessible digital injury/safety database can be developed. Therefore, relevant experimental data, including impact generation mechanisms, resulting injury assessment and provided measurements, have to be gathered, classified, digitized (when necessary), and summarized. We propose the following abstractions and categorizations to capture impact tests' various characteristics with the human body upper extremities.

A. Impact scenarios and principal test setups

The different setups for various impact experiments encountered during reviewing biomechanics literature concerning the human body upper extremities can be summarized, as shown in Fig. 2. Considering the two possible combinations of horizontal and vertical collision arrangements, four possibilities for **principal impact test setups** can be proposed by adopting similar categories to the ones introduced in [22]. The resulting categorization for arm/hand impact testing is shown in the lower part of Fig. 3. Those setups are differentiated mainly based on whether the impactor or the test subject is accelerated. Of course, both parties can be accelerating in reality simultaneously, but this is rather not implemented when designing collision experiments due to e. g. complexities in human motion tracking and data synchronization. Instead, accelerating either the impactor or the test subject is enough to regenerate/mimic the impact. Additionally, in terms of the **impact scenario**, one can distinguish between impacts where the subject is *unconstrained (U)*, *constrained (C)*, or *partially-constrained (PC)* [11]. The latter case is characterized only by a part of the subject being clamped, which is not directly in contact with the impactor, see the upper part of Fig. 3. Adding these three impact scenarios to the four principal setups results in a unique impact abstraction that is very useful and allows for a unified categorization of impact experiments employing the different mechanisms and test setups. It masks out the underlying specifics and details and enables us to make comparisons and cross-validate the findings of different experiments.

B. Impact velocity and type of impact

In terms of impact velocity, one can distinguish between *static*, *quasi-static* and *dynamic* impacts. In one of their early impact experiments, Melvin et al. classified $0.05 - 0.5$ cm/s as quasi-static velocities and $3 - 5$ m/s as dynamic impacts

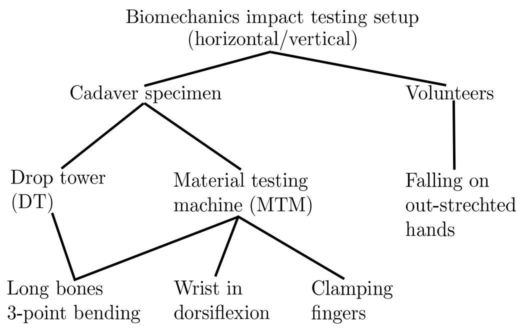


Fig. 2: Various impact testing setups encountered in the reviewed biomechanics literature for human body upper extremities.

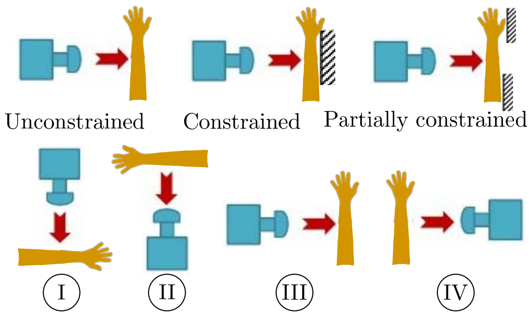
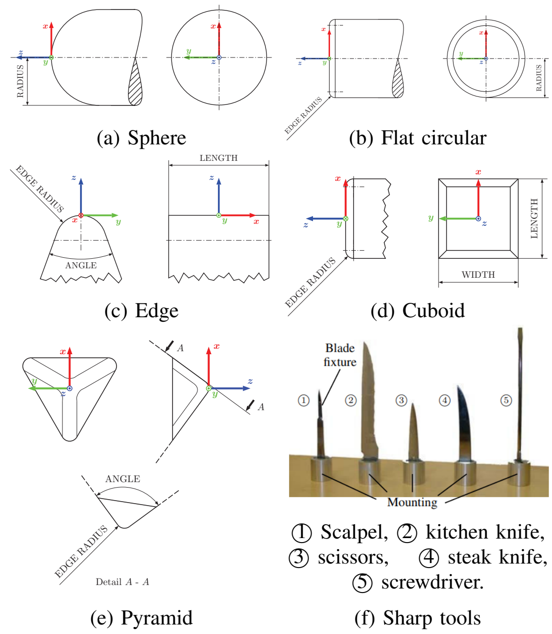


Fig. 3: Standard hand/arm impact scenarios encountered in biomechanics and principal test setups abstraction for impact experiments.

[24]. On the other hand, Bilo et al. [25] defined static loading as a relatively slow impact of forces exerted over a protracted period (>200 ms), which occurs when the skull is squeezed and compressed. Moreover, the authors attributed dynamic (or rapid) loading with the case when the impact of forces is exerted over a shorter period (<200 ms, often even less than 50 ms). Though, to the best of the authors' knowledge, there is no well-defined distinction between the three impact velocity ranges.

C. Impactor characterization

In single part experiments, sometimes the actual part of interest is tested, e. g. a car door handle in [26]. However, general geometric shapes such as cylinders or circular plates were usually used in the automotive industry to deliver precise impacts on the subject body's desired location. Those impactors vary significantly in size and shape among various impact experiments and test setups. Nonetheless, principal geometric primitives can be identified. We suggest the following minimal set for main impact primitives with their parameters, as depicted in Fig. 4. Besides the geometrical shape and size, each impactor is characterized by its mass and sometimes by distinct elastic properties such as rubber padding or any other energy-absorbing material [27]. Impactor arrangements are usually adjusted to find the injury tolerance of the respective human body parts targeted in the impact experiment. This is achieved by selecting the impactor mass and velocity values such that a particular type of injury is produced with high probability.



(a) Sphere (b) Flat circular (c) Edge (d) Cuboid (e) Pyramid (f) Sharp tools

Fig. 4: Typical impactor primitives with the according parameters. The z-axis of the coordinate frame associated to each primitive defines the impact direction $\mathbf{u}$ [14], [28].

IV. Literature Findings and Discussions

Up to now, most of the work that has been carried out regarding safe pHRI in general, and that related to injuries of the human body upper extremities in particular, was mainly driven by the automotive industry. That was based on injury tolerance information and data accumulated in the automotive field when considering traffic-accident injuries [29]–[31], or simulations built using collision models relying on them [32]. To study the pathophysiological impact response and characterize the sequel following the resulting injury, given the subinjury threshold limitations on human volunteer testing, the only viable testing surrogate is the animal testing [16]. Nevertheless, although injury tolerances of different human body part have been studied by testing with animals over abroad span of the phylogenic order, many precautions related to differences anatomy, physiology and scaling has to be respected while interpreting the data [33].

However, milder injuries and lower energies than in automobile collisions are needed for more accurate safety analysis and quantification of interaction force levels for pHRI applications [4]. Hence, generating such missing, representative collision data is a growing necessity in the robotics community. Additionally, achieving a better understanding of injury mechanisms and significant factors affecting safety in robotics requires large amounts of ground-truth, medically classified biomechanical information. For this purpose, an extensive literature review process to capture available biomechanical data from various biomechanics, sports medicine, and forensics sources alongside their experimental settings and impact characterizations has been carried out. The results of reviewed work on experimental injury biomechanics of the human body upper extremities are summarized here. We strive to represent collision data in a unified way and develop a database comprising this useful injury/safety data. We follow our data-driven approach to achieve this, where we relate robot and human input parameters to resulting injury. Acquiring adequate quantitative and qualitative knowledge is required to understand better how the effective impactor mass, relative velocity, and contact geometry (i. e. instantaneous collision data) affect the resulting injury due to human-robot impacts [14].

An overview of the results of twenty-five data sources within our massive list of reviewed references for biomechanics impact data is given Fig. 5. We note here that the reported human arm/hand injuries are mostly fractures, which can results from collisions with bulk robots or near singular configurations, enduring a (quasi-)static loading on the human upper extremity or during a secondary following impact after the initial crash. However, injuries milder than bone fractures, which may provide criteria suitable to the more light-weight robots, requires more collective effort to carry out further specific experimental investigations. Regardless of the different experimental conditions, a trend for each body part is visible. The long bones' experiments show that the necessary fracture force for all of them is in the same region. In [34], it was also concluded that the force onset direction, Lateral-Medial (LM) or Anterior-Posterior (AP) direction, is not relevant for a single bone, e. g. the humerus in this case. Regarding the Colles fracture of the wrist, there is no clear threshold resulting from Fig. 5. Most of the destructive cadaver tests indicate a fracture force threshold in the same region as for the long bones. However, the forces occurring in real impact testing setups with volunteers seem to overlap with the fracture region. One reason for this effect might be that muscle tension and other soft tissues stabilize the bones. Therefore, the real fracture tolerance seems to be higher than indicated by the cadaver specimen tests. For fingers in the closing car window arrangement, the volunteers applied much lower forces than the fracture tests suggests. This contrasts with the wrist results and indicates that the fingers' sensitivity is a lot higher than for the wrist.

Apart from the conclusion of unclear fracture threshold values, another approach would be to use risk-functions for fractures like presented in [60] and [61]. These functions relate a certain impact force or bending moment to a probability of a fracture. In [61], Weibull distributions were used for the impact forces on the metacarpophalangeal joint (MCP) and the proximal interphalangeal joint (PIP) joint, see the last row in Fig. 6. Duma et al. [60] used logistic regression functions for the humerus, elbow, forearm, and wrist. In that paper, the test data originated from a small female specimen and, therefore, their results represent a conservative estimate only. For the long bones, the bending moment is related to fracture probability, and for the joints, the axial force is used, see first and second rows of Fig. 6. Several studies also report bending moments for the long bones. These are shown commonly as a table in books like [62] and see Tab. I. From these values, it is visible that the female bones' bending moment is lower than of male bones. As already concluded by [34] the scaling method introduced by [63] seems to result in too high values. This can also be seen in the table by comparing the scaled with the unscaled results.

In terms of robotics, only one recent study by Povse et al. [19] is available. The authors treated the systematic injury analysis of lower arm robot-human impacts by developing a passive mechanical lower arm (PMLA) that mimics the human impact response and enables prediction of mild contusions and lacerations. The PMLA biofidelity was verified by a number of comparative impact experiments with a human volunteer, where the respective dynamic impact responses in terms of impact forces and impact energy densities (as functions of end-effector velocity) showed very good consistency. Two different robots were used for performing the impact experiments, whereas the elbow joint was controlled to mimic the human elbow impedance behavior. The acquired experimental impact data-sets were encapsulated into a robot independent safety curve, see Eq. 1. Overall, the injury estimation using their developed PMLA device showed a good agreement with the experimental results of unconstrained lower arm collisions, specifically for hematoma and lacerations. The PMLA impact results are not repeated here since the focus is more on the unclassified data from biomechanics research and other literature.

Bending moment		Reference
Male [Nm]	**Female** [Nm]	
115	73	[64]
151	85	[37]
157	84	[34]
230	130	[34], scaled
138		[35]
	154	[65]
217	128	[58], scaled

TABLE I: Maximum bending moments of a human humerus according to different studies, adapted from [62]. Some values are "scaled" to 50%ile male and 5%ile female [63].

Due to the lack of robotics relevant data-sets for milder hand injuries, we further extended our search into standard tolerance levels from similar use cases in other industrial domains. As a result, we found some safety requirements covering the manufacturing machines for shoes and leather goods [66]. Specifically, a protection device for the worker's fingers that divides the machine closing movement into two phases was described. The standard defines a force limit of $150\,\mathrm{N}$ or contact pressure limit of $50\,\mathrm{N/cm^2}$, respectively, for performing the first phase, while in the second phase the operating force can be used.

V. Conclusions and Future Work

An extensive literature review process to capture available data from relevant sources alongside their impact characterization, collision scenarios, and testing setups has been completed for the human body upper extremities. The results of this work can be used as a comprehensive guideline for both the injury biomechanics and robotics communities when designing impact experiments and studying physical interaction between robots and human body upper extremities. Nonetheless, one limitation that was obvious after our extensive literature reviews is that milder injuries for robots are not extensively investigated. For this reason and based on the presented results' insights, anthropometric dummy crash-testing experiments against a robot arm are planned as future work to generate more robotics-relevant impact data-sets.

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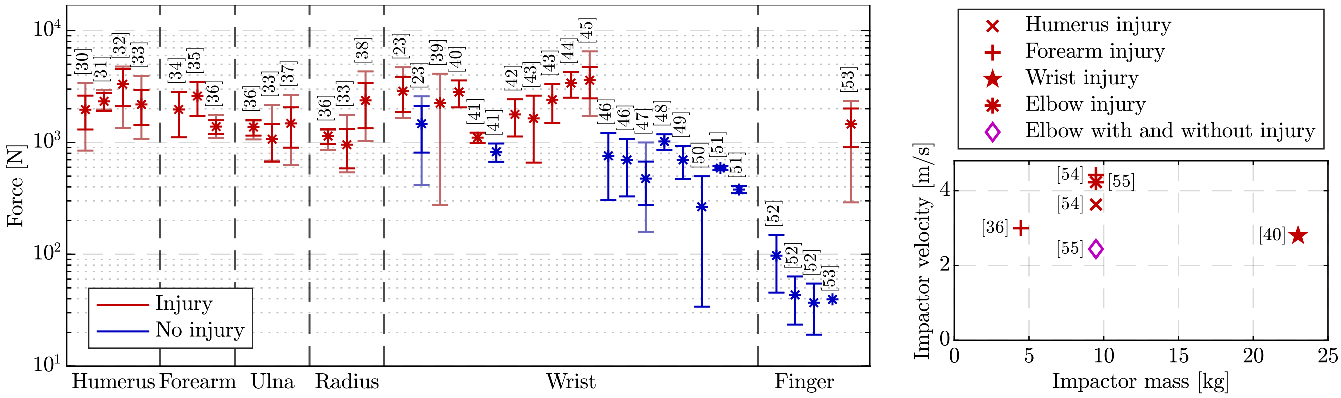

Fig. 5: On the left side, a synopsis plot of the force values reported in twenty-five reviewed data sources, that mostly investigate fracture injuries, is given. Red color means that fractures occurred and blue means no injuries, e. g. while tests with volunteers. The asterisks mark the mean values, the solid lines represent the standard deviation and the light colored lines depict the range between lowest and highest values. Humerus, forearm, ulna and radius are tested in three-point bending tests, wrist means the human falling on hands setup, and finger denotes the jamming in closing car window arrangement. On the right side, the mass-velocity plot shows the results of drop tower experiments. Red colored marks mean that in all tests injuries occurred at this mass-velocity combination of the impactor. If both, fractures and no injuries, were resulting the mark is colored magenta. Humerus and forearm are results from three-point bending tests, wrist means the human falling on hands setup, and elbow is similar to three-point bending tests, but with overextending the elbow joint.

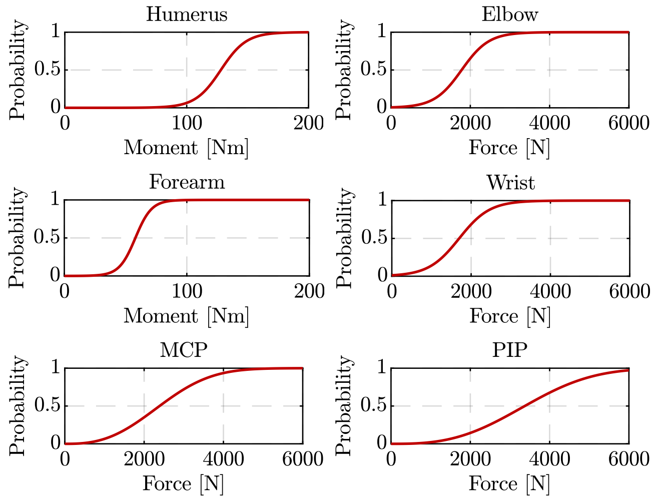

Fig. 6: Probability functions for fractures of the arm adapted from [60] and of the hand adapted from [61].

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