# Development of a Physical and Mathematical Ballistic Skin Simulant

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**Abstract.** Ideally, non-lethal weapons (NLWs) incapacitate or repel with a low probability of fatality or permanent injury. Rapid NLW market growth and a growing number of new kinetic weapons necessitate further relevant skin injury biomechanics research. Unfortunately, NLW development and evaluation is limited by a lack of universally accepted standard test protocols or materials. To fill this need, Luna Innovations developed TrueSkin™, a physical simulant of human skin, with a coupled Finite Element Mathematical (FEM) model of the material. (U.S. Army SBIR Phase II contract W91CRB-17-C-0032). TrueSkin comprises a proprietary nanofiber-reinforced hydrogel that mimics the architecture of human skin extracellular matrix and ultimately the skin biomechanics to failure and response to non-lethal munition impact. The physical skin simulant is designed to improve the reliability of injury risk evaluation for less-than-lethal ballistic projectiles and was iteratively designed and tested to match the material properties (ultimate tensile strength, stretch at failure) of postmortem human skin tissue. To accomplish this, the TrueSkin physical simulant was evaluated at relevant dynamic strain rates using custom equipment and protocols developed at the University of Virginia Center for Applied Biomechanics (UVA-CAB) for testing human skin tissue. Independent evaluation at HP White Laboratories validated penetration resistance of TrueSkin against established postmortem human subject (PMHS) penetration data using similar 12 gauge fin-stabilized rubber projectiles. The promising results from mechanical characterization of laboratory specimens demonstrate material property simulation capability, and comparison of penetration resistance with existing PMHS data provides initial validation of ballistic response.

## 1. INTRODUCTION

Non-lethal weapons (NLWs), including less-lethal kinetic energy (KE) impact munitions, are designed to incapacitate or repel targets with a low probability of fatality or permanent injury. These specialty impact munitions are typically deployed during encounters with aggressive subjects where a less-than-lethal response is needed to minimize the potential for significant injury to the target or others nearby. Single projectile KE impact munitions are typically utilized as their accuracy enables direct fire towards a specific area of the body to inflict blunt trauma with minimal risk of serious injury. Rapid market growth has created an intensely competitive effort to develop and commercialize more effective and safer NLWs. Impact munitions, such as rubber projectiles, can serve as valuable tools to military and law enforcement agencies worldwide[1]. However, concern remains regarding adverse effects that the munitions can have on potential targets when penetration through the skin occurs[2].

According to a report by Haar et al, over 2000 penetrating injuries, ranging from minor to significant, and over 50 deaths from impact munitions have occurred between the years of 1990 and 2007[3], which could increase as new projectiles are rapidly developed. Recent studies evaluated almost 1000 deployments of less-lethal kinetic energy rounds, primarily focused on 12-guage bean-bag and plastic baton rounds and found that over 80% of these deployments result in injury [4]. While most injury were contusions (51%) and abrasions (31%), fractures and other penetrating injuries were reported in these smaller caliber KE impact munitions.

As a result of these case reports potential to cause penetrating injury to the body, thorough human effects, or safety assessments, must be accomplished to define minimum engagement distances, maximum velocities, and to assess the risk of significant blunt trauma[5], [6] and projectile penetration across entire operation ranges. Numerous experimental and computational models have been developed to predict the risk of injury due to rigid projectile impacts, and many previous studies focused on safety evaluation have been used as a guide in the early stages of less-lethal projectile design[1], [7]. Unfortunately, NLW safety research and evaluation standards are limited by a lack of universally accepted test protocols or materials to determine the risk of significant injury of non-penetrating NLWs with statistical rigor to ensure safe use.


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The ability to test NLW safety in a controlled environment is paramount to enable a thorough biomechanical assessment of impact prior to use in the field. The risk of penetrating trauma is especially important to evaluate during safety assessments of NLWs due to the dramatic increase in injury severity that can occur when the munition penetrates into the body cavity[8]. Current methods for evaluating the injury risk, and primarily the penetration risk of NLWs, are limited. Various ballistic injury surrogates have been utilized including soap, gelatin, clay, animals, PMHS and other materials [9], [10]. Although many of these materials can be used to determine energy transfer from the munition to the tissue surrogate, they lack the visco-elastic nature of human tissue and are primarily adapted from previous methods to evaluate lethal, penetrating ballistics.

A previous study by Bir et. al, using PMHS, assessed the skin penetration of a 12 gauge, fin-stabilized, rubber projectile [2]. In this study, un-embalmed PMHS were impacted at relevant velocities by the KE impact munitions and for each impact, the energy density (Ea) was calculated and injury was determined by evaluating the underlying tissue damage. Chamois skin was also used as part of skin surrogate's development for skin penetration assessment and showed acceptable biofidelic results, however, its thickness inconsistency and reproducibility reduce its practicality [8], [11]. Efforts have been made to develop affordable synthetic skin simulant (silicone and urethane) with a stable shelf life to exhibit biomechanical failure equivalence to human skin also show little success [12].

Human dermis is a matrix of collagen (approximately 35 vol%) and elastin fibers (approximately 0.4 vol%) that are interwoven in a highly hydrated proteoglycan gel. The dermis is capable of withstanding large deformations and the strength of the skin is primarily attributed to collagen fibers, which are almost inextensible and fail at strains of 5-6% and strengths of 147-343 MPa[13]. Elastin, on the other hand, is highly deformable, and provides the dermis with its high elasticity. For accurate simulation of force loading and response to NLW impact, we pursued skin simulant methods that would recreate these strong and elastic components and simulate the tough material properties of human skin.

Using parallel biomechanical characterization of PMHS skin, the current study was conducted to develop a biomimetic human skin surrogate for use in the safety evaluation of less-lethal KE impact munitions. In order to achieve similar penetration resistance, a custom hydrated nanofiber material was devised and the tensile mechanical properties to failure were determined and compared to human skin (PMHS). Re-evaluation of a previous NLW penetration study using PMHS was accomplished to determine the penetration resistance of a 12 gauge KE impact munition over soft issue in an effort to validate the resistance of the skin surrogate developed in this study. Finally, a penetration resistance study was accomplished using the newly developed surrogate and similar munitions for direct comparison to the PMHS penetration resistance. Identification of a repeatable, cost-effective, biomimetic surrogate of human tissue to evaluate KE impact munitions will be a valuable too for munitions manufacturers during development, law enforcement agencies during munition acquisition activities, and government agencies during test standard development.

## 2. METHODS

### 2.1 PMHS and Skin Surrogate Mechanical Testing

Skin samples for mechanical characterization were excised from the back of six male PMHS at the Center for Applied Biomechanics, University of Virginia (UVA), USA. All test procedures were approved by the UVA Institutional Review Board prior to any testing and the PMHS were screened for pre-existing pathologies to avoid skin diseases that may affect skin quality. The PMHS represent an average of 57±11 years old, 178.6±3.8 cm in height and weighed 88.4±19.8 kg adult male (Table 1).

**Table 1.** Post-mortem human subject information (Mechanical Characterization)

| Specimen ID | Age (years) | Height (cm) | Weight(kg) |
|---|---|---|---|
| 795 | 60 | 175.3 | 83.9 |
| 757 | 49 | 185.4 | 122 |
| 702 | 42 | 178 | 86 |
| 733 | 74 | 180.3 | 78.9 |
| 919 | 59 | 177.8 | 96.6 |
| 680 | 58 | 175 | 63 |

All the PMHS were thawed for 72 hours at room temperature prior to skin excision for testing. The orientation of the skin sample was determined with reference to anatomy illustration of Langer line[14], and both parallel (0°) and perpendicular (90°) samples with respect to the Langer line were included for testing (Figure 1, right). For each PMHS, 2 different sizes of skin samples were excised on the left side

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of the back for uniaxial tensile to failure (static and dynamic) and stress relaxation tests. Each skin sample thickness was measured using a digital caliper prior to testing and the measured mean thickness was 3.33±0.87 mm. Further detail regarding stress relaxation testing utilized for constitutive modeling of human skin biomechanics can be found in Kong et. al [15].

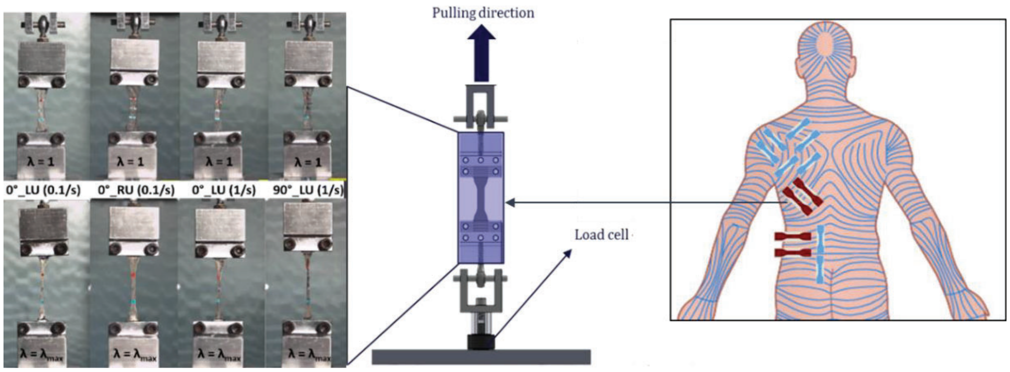


**Figure 1.** Overview of Tensile Testing of PMHS.

The static (1/s) tests were performed using the Instron Model 8874 servohydraulic actuated test machine (Instron, Canton, MA) and a custom-built gravity-based drop tower was used for the DT tests. Dogbone sample was clamped with 80 grit sandpapers at both ends of the custom test fixture (Figure 1, middle) to avoid slippage during testing. The test was initiated by moving the crosshead of the machine until skin failure occurred. A 1000 lbf (4.4 kN) Honeywell model 31 piezoresistive load cell (Honeywell, Charlotte, NC) was used to measure the force at 1000 and 10,000 sampling rates for ST and DT tests respectively. Similarly, test videos were recorded at 1000 fps for ST tests and 10,000 fps for DT tests using a Memrecam GX-1 high-speed camera (NAC Image Technology, Simi Valley, CA). Sharpie marks which were placed *in situ* were used for determining stretch values through video tracking software (Figure 1, left) (Tracker, ver. 4.11.0). A trigger box was utilized to activate data acquisition of the load cell and video recording simultaneously when the test was initiated. Engineering stress-stretch curve was constructed based on the measured force and the video tracked displacement. The engineering stress was calculated by dividing the measured force by the undeformed cross-section area. The stretch ratio (λ) was defined as the ratio between the current gauge length and original gauge length.

**2.2 Human Skin Surrogate**

Two different composite approaches were taken in an attempt to simulate human skin microstructure and response to applied loads (Figure 2). The first involved rayon microfiber-reinforcement of commercially available silicone. Early development efforts focused on recreating the fiber loading and failure mechanisms in this simulant approach (Figure 2, white box). Preliminary mechanical testing (data not shown) suggested that the microfibers were adversely impacting the mechanical properties of the samples because of nonuniform distribution and poor adhesion to silicone matrices. We transitioned surrogate development to recreation of the nanofiber structure and high hydration of human skin for optimal recreation of the viscoelasticity that dictates the mechanical properties and penetration resistance of skin. The final simulant approach therefore comprised Kevlar nanofiber (KNF) [16] reinforced poly(vinyl alcohol) hydrogel [17] (Figure 2,right). All mechanical testing (method describe above) and penetration resistance testing utilized the KNF surrogate.

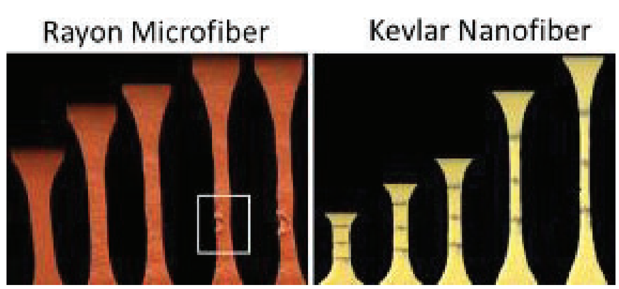


**Figure 2.** Overview of Skin Surrogates Evaluated.


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### 2.3 Statistical Analysis of PMHS Penetration

A previous report by Bir et al [2] contained the results of ballistic testing of 8 PMHS against a 12-gauge, fin-stabilized rubber round (Figure 3). The eight cadaveric specimens, four male and four female, were procured from either the Wayne State University (WSU) Body Bequest Program or the University of Michigan (UM) Anatomical Donations Program. The cadavers were fresh, not embalmed. A summary of each cadaveric specimen is located in Table 2.

**Figure 3**. 12-gauge rubber round and cartridge.

**Table 2.** Post-mortem human subject information (Penetration Resistance)

| Cadaver | | | Age, | Height | | | Weight | | |
|---|---|---|---|---|---|---|---|---|---|
| ID | Source | Sex | yrs | cm | ft | in | kg | lbs | BMI |
| 31222 | UM | F | 58 | 162.5 | 5.331 | 64.0 | 71.21 | 157.0 | 27.0 |
| 31234 | UM | M | 58 | 175.5 | 5.758 | 69.1 | 57.61 | 127.0 | 18.7 |
| 562 | WSU | F | 72 | 168.0 | 5.512 | 66.1 | 72.57 | 160.0 | 25.7 |
| 430 | WSU | M | 75 | 174.0 | 5.709 | 68.5 | 84.14 | 185.5 | 27.8 |
| 31155 | UM | M | 76 | 174.0 | 5.709 | 68.5 | 73.03 | 161.0 | 24.1 |
| 31480 | UM | M | 77 | 172.0 | 5.643 | 67.7 | 78.02 | 172.0 | 26.4 |
| 545 | WSU | F | 78 | 155.0 | 5.085 | 61.0 | 52.16 | 115.0 | 21.7 |
| 563 | WSU | F | 80 | 164.0 | 5.381 | 64.6 | 68.04 | 150.0 | 25.3 |

Each PMHS sustained a maximum of 25 impacts consisting of shots to the anterior and posterior thorax, abdomen, and legs, for a total of 158 total impacts. The 10 specific locations of impact include 5 areas where bone lie directly under the skin (sternum, anterior on rib, anterior between ribs, posterior on ribs, scapula) and 5 fleshy areas either devoid of bone or with muscle/fatty layer before bone (liver, lateral to umbilicus (belly), posterior lower back, proximal femur, distal femur). Following each impact to a given location, a visual inspection of the injury was performed, and the wound was labeled penetrating or non-penetrating. Penetrating wounds were determined as such by evaluating whether the impactor disrupted not only the skin, but underlying tissue such as subcutaneous fat and/or muscle. Slight tearing (laceration), discoloration or marking of the skin without damage to underlying tissue was recorded and regarded as non-penetrating. Projectile velocity was measured with a single chronograph placed 22 in. from the PMHS. The mass and diameter of each round was measured and the energy density (J/cm$^2$) was mathematically determined.

As a validation of the skin simulant prototype, it was desired to replicate this PMHS dataset as closely as possible to see if the skin simulant would produce results similar to actual human skin. As the majority of methods for penetration currently use gelatin as backing material to simulate muscle/soft flesh [8], only the soft-backed PMHS impact locations were utilized to derive "actual human skin" expected performance, or truth set, for initial comparison and validation of the synthetic skin material. The 5 soft-tissue body locations from 8 PMHS totaled 68 shots. The data for these are plotted in Figure 4 with non-penetrations plotted as 0 and penetrations plotted as 1.

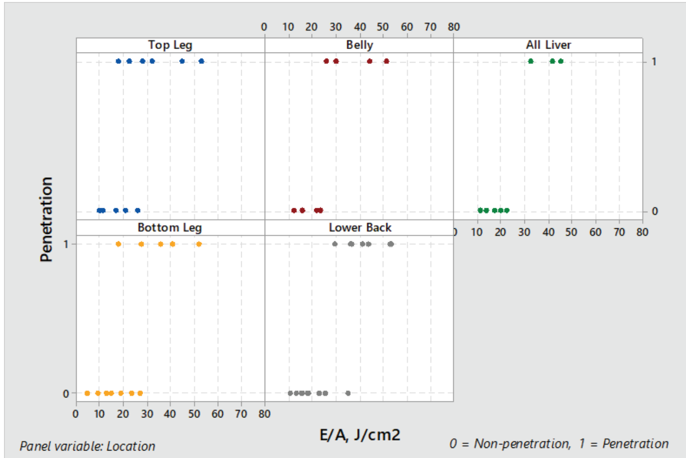


**Figure 4**. All data from 5 soft tissue backed locations.

To see if the five body parts produced similar results, or data from the same population, and therefore could be combined, the different populations statistical test was performed. The 0.10 alpha (or (1-alpha) = 0.90 difference confidence) was used to determine the level of significant difference. Parameter estimates using likelihood ratio analysis were calculated using three different link functions (Log Normal, Probit, Logit). A plot of the log-likelihood response curve along with lower and upper 90% confidence bounds were calculated. The log-normal link function is preferred since, unlike a probit or logit, it does not produce any negative stimulus values for the response or confidence curves.

### 2.4 Skin Simulant Penetration Testing

Independent penetration testing was conducted on an indoor range at HP White Laboratories (Street, Md) at ambient conditions utilizing a custom pneumatic test setup to control the KE projectile velocity. Testing was conducted using a 12 gauge less lethal rubber rocket manufactured by Defense Tech (DT3021). The test platform (Figure 5) consisted of a 20% ballistic gelatin covered by the skin surrogate, secured with a custom outer clamping ring to improve consistency of boundary conditions under applied load (impact munition force).

**Figure 5**. Skin Surrogate and Ballistic Gelatin utilized for penetration testing.

The test samples (Figure 5) were positioned 6.17 feet from the muzzle of the custom barrel to produce zero (0°) degree obliquity impacts (Figure 6). Each projectile was weighed prior to testing. High speed recording at 10,000 frames/sec was used to determine the striking velocity of the projectile. Final injury response criteria were selected and included: *(1)* No Injury – neither the skin simulant nor the gel block indicated irreversible damage, *(2)* Contusion – the gel block showed indication of impact but the skin simulant was not perforated, *(3)* Laceration – the skin simulant was perforated as indicated by visible light, and *(4)* Complete Penetration – the underlying gel block surface was fractured. As a validation of the skin simulant prototype, it was desired to replicate the Bir 2005 cadaveric experiment [2] as closely as possible to see if the skin simulant would produce results similar to actual human skin.

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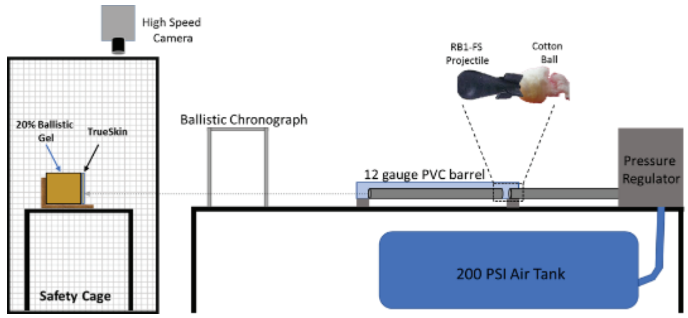


**Figure 6**. Overview of Penetration Testing of 12 gauge KE impact munition with Skin Surrogate.

Energy density (Ea) was calculated using projectile mass (~6g), face area(2.45 cm$^2$) and impact velocity. After each shot, the skin surrogate was removed from the ballistic gelatin and both the skin surrogate and the underlying gelatin were examined. The horizontal yaw was captured using a high-speed video camera mounted above the simulant. The depth of the projectile was also captured by the high-speed video camera with parallax effects accounted for in the gelatin.

## 3. RESULTS

### 3.1 Mechanical Characterization

The uniaxial tensile test matrix and average responses of the uniaxial tensile PMHS skin samples relevant to the development of a skin surrogate is summarized in Table 3. Among the 48 skin samples tested at the UVA-CAB, 5 samples slipped out of the test fixture and were permanently deformed during testing, therefore, these samples were excluded from data analysis. Additional details regarding mechanical characterization of these PMHS skin samples can be found in Kong et al [15].

**Table 3.** Average uniaxial tensile test results of human skin

| Orientation | Strain rate (/s) | UTS (MPa) | $\lambda_f$ |
|---|---|---|---|
| **Parallel** | 1 (n=11) | **28.4**±6.3 | **1.76**±0.14 |
| | 75 (n=5) | **20.6**±7.8 | **1.75**±0.18 |
| | 180 (n=6) | **25.6**±5.4 | **1.75**±0.07 |
| **Perpendicular** | 1 (n=10) | **22.6**±4.6 | **1.97**±0.15 |
| | 75 (n=5) | **16.6**±5.8 | **1.78**±0.12 |
| | 180 (n=6) | **20.7**±4.9 | **1.81**±0.10 |

For the PMHS samples tested at the same strain rate, the parallel (0°-red) and perpendicular (90° - blue) stress-stretch-relationships are compared (Figure 7). The average UTS of all samples tested at all strain rates (n = 43) was 23.3 MPa and the standard deviation was 6.64 MPa. Additional efforts to perform statistical analysis between these groups was combined with the stress relaxation data (data not shown) in an effort to develop a parallel constitutive model of skin. However, the primary output of the PMHS mechanical characterization with regard to the development of an appropriate skin simulant was the UTS, stretch at failure ($\lambda_f$), and the overall stress-stretch relationship (appearance of toe region, strain-stiffening, and damage propagation) (Figure 7).

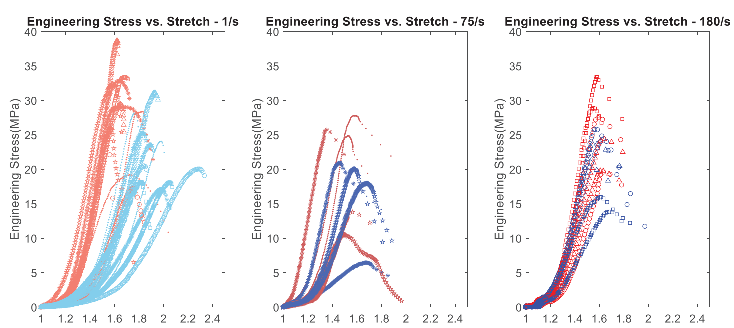


**Figure 7.** Stress-Stretch profiles of varied orientations (0°-red, 90°-blue) under strain rates (1,75,180/s)

As a result of iterative improvement of fabrication protocols and selection of appropriate concentrations of the PVA and KNF solutions and the ratio of the two composition, we fabricated custom skin surrogate prototypes (Figure 8 -left) that exhibited stress-stretch relationships (Figure 8-right) and mechanical properties within the range of PMHS mechanical data collected. Specifically, the skin surrogate exhibited an ultimate tensile strength of 26.97 ± 0.23 MPa and failed at a stretch ratio of 2.17 ± 0.01. An overview of the mechanical properties compared between PMHS and the skin surrogate can be seen in Table 4.

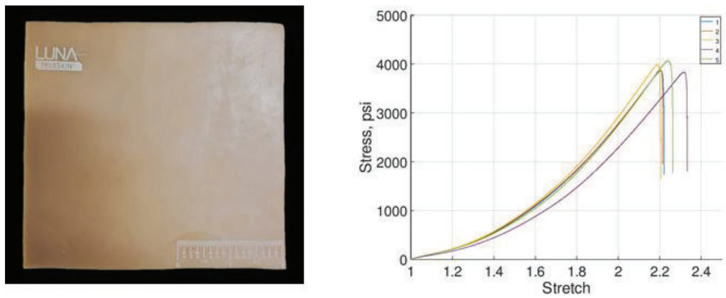


**Figure 8.** Final Skin Surrogate Prototype and Resulting Stress-Stretch relationship

**Table 4.** Average uniaxial tensile test results of human skin compared to surrogate

| Material | Ultimate Tensile Strength | Stretch at Failure |
|---|---|---|
| PMHS Skin | 23.30 ± 6.64 (MPa) | 1.81 ± 0.15 |
| Custom Nanofiber Skin Surrogate | 26.97 ± 0.23 (MPa) | 2.17 ± 0.01 |

### 3.2 PMHS Penetration Evaluation

Eight fresh cadaveric specimens, four male and four female, were tested for the Bir report [2]. Each PMHS sustained impacts consisting of shots to the anterior and posterior thorax, abdomen, and legs, for a total of 158 total impacts. The 10 specific locations of impact included 5 'hard-backed' areas where bone lies directly under the skin (sternum, anterior on rib, anterior between ribs, posterior on ribs, scapula) and 5 'soft tissue' areas either devoid of bone or with a muscle/fatty layer between skin and bone (liver, lateral to umbilicus (belly), posterior lower back, proximal femur, distal femur). Since the skin simulant evaluation uses gelatin as backing material, only the soft tissue PMHS locations were used.

The 5 soft tissue locations from 8 PMHS totaled 68 shots. To see if the five body parts produced similar results, or data from the same population, and therefore could be combined, the different populations statistical test was performed. The 0.10 α (or (1-α) = 0.90 difference confidence) was used to determine the level of significant difference and identified that the lower back and belly produced different responses (Table 5 -red).

**Table 5.** Difference Confidence for five Soft-tissue locations

| | Belly | All Liver | Bottom Leg | Top Leg |
|---|---|---|---|---|
| Lower Back | 91.2 | 43.9 | 62.1 | 84.6 |
| | Belly | 0.0 | 85.0 | 86.6 |
| | | All Liver | 60.2 | 70.5 |
| | | | Bottom Leg | 15.3 |

Deleting the Belly data resulted in 50 data points. The difference confidences are presented in Table 6, showing none exceeded the 90% difference confidence. Parameter estimates using likelihood ratio analysis were calculated using three different link functions for the combined dataset and are presented in Table 7.

**Table 6.** Difference Confidence for five Soft-tissue locations

| | All Liver | Bottom Leg | Top Leg |
|---|---|---|---|
| Lower Back | 43.9 | 62.1 | 84.6 |
| | All Liver | 60.2 | 70.5 |
| | | Bottom Leg | 15.3 |

**Table 7.** Parameter Estimates from four soft-tissue locations combined

| N | Link Function | Mu | sigma | 10% | 90% | L90CL Mu | U90CL Mu | L90CL sig | U90CL sig |
|---|---|---|---|---|---|---|---|---|---|
| 50 | Log Normal | 25.43 | 0.2883 | 17.57 | 36.80 | 22.44 | 29.05 | 0.1968 | 0.4476 |
| | Probit | 26.57 | 7.13 | 17.43 | 35.70 | 23.53 | 30.23 | 4.84 | 11.11 |
| | Logit | 26.55 | 7.36 | 17.21 | -- | 23.45 | 30.24 | 4.72 | 11.96 |


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A plot of the log-likelihood response curve along with lower and upper 90% confidence bounds are presented in Figure 9. The log-normal link function is preferred since, unlike the other two, it does not produce any negative stimulus values for the response or confidence curves. Per Table 7 and Figure 9, in order for the skin simulant to be considered a match to the soft-tissue backed PMHS data, the surrogate should produce a mean energy density (Ea50) between 22.4 and 29.0 J/cm$^2$.

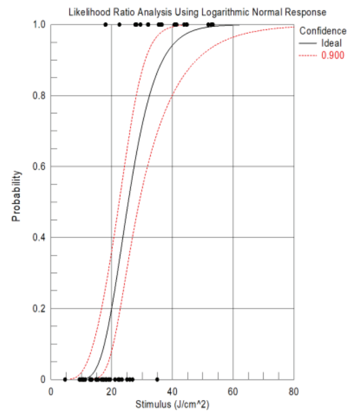


**Figure 9.** Response Curve and 90% Confidence bounds for soft-tissue backed PMHS locations

### 3.3 Surrogate Penetration Evaluation

Figure 10 displays the visual responses for skin simulant and the underlying ballistic gelatin from the first 15 shots, along with their corresponding projectile velocity and classification. 40 shots were accomplished in this study, of which 14 exhibited either high yaw (>20°) or struck the testing frame and were therefore not included into analysis. As a result, the penetration resistance of the remaining 26 shots was accomplished for comparison to the PMHS dataset (Figure 9).

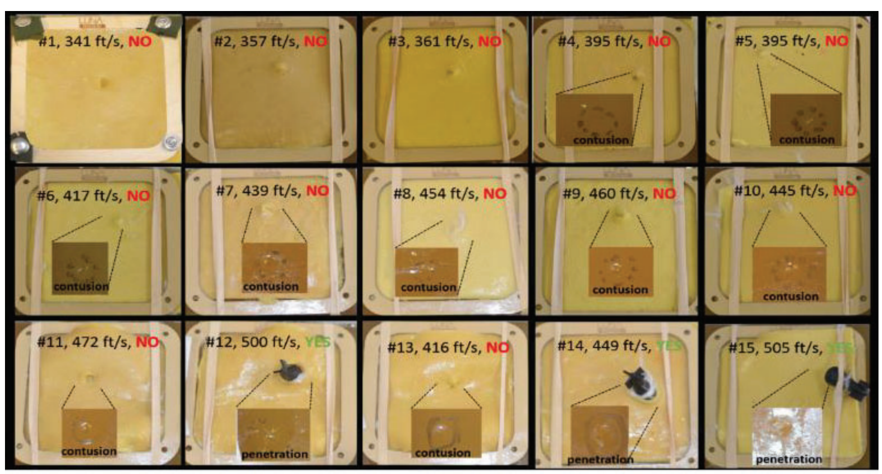


**Figure 10.** Visual Examination Results from Penetration Testing at HP White Laboratories

A log-normal analysis was utilized along with the likelihood ratio to determine 90% confidence intervals for the probability of penetration as it relates to the average energy per unit area, following methods utilized for PMHS testing (Figure 11). The skin simulant produced a mean energy density (Ea50) of 22.9 J/cm$^2$. Figure 8 presents comparison of the results from the PMHS-defined injury risk 'truth-set' and the surrogate penetration testing results accomplished at HP White Laboratories.

**Table 8.** Comparison of Ea50 for Soft-Tissue Backed PMHS and Skin Surrogate

| Prototype | Ea50 Confidence Interval (Average) [J/cm$^2$] |
|---|---|
| Re-analyzed PMHS data | 22.44-29.05 **(25.43)** |
| **TrueSkin Prototype** | 20.71-24.75 **(22.86)** |

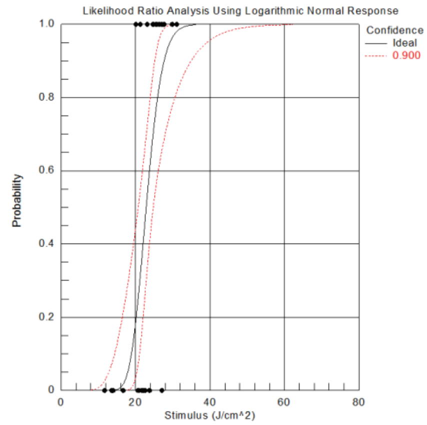


**Figure 11.** Response Curve and 90% Confidence bounds for Surrogate testing

## 4. DISCUSSION AND CONCLUSION

### 4.1 Discussion

A synthetic skin simulant has been developed in an effort to fulfill a need from kinetic munition developers and safety evaluators. This need simulant and set of test methods was accomplished by precisely characterizing the mechanical properties of human skin, tuning the composition of a hydrated nanofiber material, and evaluating the penetration resistance of the surrogate to 12 gauge rubber projectiles. In this study, PMHS skin samples were investigated under different strain rates and skin orientations and were utilized to generate stress-stretch profiles of human skin to failure. This data was utilized to (1) iteratively inform parallel development of a fiber-reinforced hydrated skin model by setting critical mechanical properties (UTS and $\lambda_f$) and (2) develop a constitutive model of human skin for future use in finite element modeling. While many groups have focused on characterizing the mechanical properties of human skin[18], [19], these are typically accomplished without testing to failure, a critical need to define the properties relevant to NLW testing and evaluation. Additionally, many of these studies utilize PMHS samples with excessive age[13].

While previous studies have examined the failure mechanics of animal skins, such as goat or porcine[20], [21], to failure these natural materials are known to vary from human skin biomechanics. The results of this study demonstrate how mechanical characterization of human skin (PMHS) can be accomplished to drive the design of a surrogate material that matches relevant mechanical properties, with higher reliability (lower standard deviation) and without the costs and logistical burdens associated with human subject testing.

More recent studies have attempted to standardize the evaluation of NLWs, including KE impact munitions, with the NATO Standardization Recommendation (STANREC)[22]. These recommendations include the use of test methods focused on assessing skin penetration in addition to the blunt impact effect. While numerous studies have examined the relationship between applied energy (Ea) and risk for significant injury using surrogate materials, these studies have primarily been focused on simulating and investigating viscous or blunt impact effects[3], [23]. Studies focused on penetration testing often utilize synthetic or natural materials for representing skin, such as Chamois[8]. However, these studies either utilize natural materials which can exhibit variation on par with human subjects, or utilize materials which have not been verified to match biomechanical properties of human skin.

### 4.2 Conclusion

The current study investigated relevant mechanical properties of human skin which was used to inform a custom hydrated-nanofiber skin simulant. The synthetic skin simulant was then utilized in a kinetic munition penetration evaluation protocol with a ballistic gelatin backing. The penetration resistance of a 12 gauge rubber projectile was determined and compared to previous PMHS penetration data [2]. The following statements highlight the takeaways from this study:

- Definition of the mechanical properties of human skin relevant to NLW penetration resistance (UTS, λ at failure) with precise characterization of PMHS evaluated at the UVA-CAB.
- Fabrication of a proprietary hydrated nanofiber skin simulant for accurate simulation of human skin mechanical properties.


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- Evaluation of previous PMHS penetration resistance dataset and establishment of penetration likelihood over soft-tissue backed anatomies.
- Validation of physical simulant penetration performance with independent testing at HP White Laboratories.

With further refinement of test methods for penetration evaluation and validation of the simulant and methods developed in this study for additional KE impact munitions, the surrogate can provide a valuable resource for key stakeholders in the development and safety evaluation of NLWs. Data generated from this model can eventually reduce the cost and logistic burden of PMHS or animal testing and can be used to generate safety profiles of NLW or other projectiles with higher statistical confidence.

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**References**

[1] J. A. Kapeles and C. A. Bir, “Human Effects Assessment of 40-mm Nonlethal Impact Munitions,” Hum. Factors Mech. Eng. Def. Saf., vol. 3, no. 1, 2019.
[2] C. A. Bir, S. J. Stewart, and M. Wilhelm, “Skin Penetration Assessment of Less Lethal Kinetic Energy Munitions,” J. Forensic Sci., vol. 50, no. 6, pp. 1–4, 2005.
[3] R. J. Haar, et al, “Death, injury and disability from kinetic impact projectiles in crowd-control settings: A systematic review,” BMJ Open, vol. 7, no. 12. 2017.
[4] K. Hubbs and D. Klinger, “Impact munitions: Data base of use and effects,” Ncj 204433, 2004.
[5] J. Pavier, et al, “On ballistic parameters of less lethal projectiles influencing the severity of thoracic blunt impacts,” Comput. Methods Biomech. Biomed. Engin., vol. 18, , 2015.
[6] C. Bir and D. C. Viano, “Design and injury assessment criteria for blunt ballistic impacts,” J. Trauma - Inj. Infect. Crit. Care, vol. 57, no. 6, pp. 1218–1224, 2004.
[7] C. Bir, D. Viano, and A. King, “Development of biomechanical response corridors of the thorax to blunt ballistic impacts,” J. Biomech., vol. 37, no. 1, pp. 73–79, 2004.
[8] C. A. Bir, M. Resslar, and S. Stewart, “Skin penetration surrogate for the evaluation of less lethal kinetic energy munitions,” Forensic Sci. Int., vol. 220, no. 1–3, pp. 126–129, 2012.
[9] A. Chanda, et al, “Experimental study on tissue phantoms to understand the effect of injury and suturing on human skin mechanical properties,” Proc. Inst. Mech. Eng. 2017.
[10] M. Xiong, B. Qin, S. Wang, R. Han, and L. Zang, “Experimental impacts of less lethal rubber spheres on a skin-fat-muscle model,” J. Forensic Leg. Med., vol. 67, pp. 7–14, 2019.
[11] A. Papy, et al, “Definition of a standardized skin penetration surrogate for blunt impacts,” in 2012 IRCOBI Conference Proceedings - 2012, pp. 486–493.
[12] “Mechanical Characterization of Soft Tissue Simulant Materials,” Adv. Exp. Mech., 2017.
[13] M. Ottenio, et al, “Strain rate and anisotropy effects on the tensile failure characteristics of human skin,” J. Mech. Behav. Biomed. Mater., vol. 41, pp. 241–250, 2015.
[14] A. N. Annaidh, et al, “Mechanical properties of excised human skin,” in IFMBE Proceedings, 2010, vol. 31 IFMBE, pp. 1000–1003.
[15] M. B. Panzer, K. Kong, et al, “Dynamic Mechanical Properties of Human Skin,” in PASS, 2020.
[16] J. Lin, S. H. Bang, M. H. Malakooti, and H. A. Sodano, “Isolation of Aramid Nanofibers for High Strength and Toughness Polymer Nanocomposites,” ACS Appl. Mater. Interfaces, 2017.
[17] S. R. Stauffer and N. A. Peppast, “Poly(vinyl alcohol) hydrogels prepared by freezing-thawing cyclic processing,” Polymer (Guildf)., vol. 33, no. 18, pp. 3932–3936, 1992.
[18] A. J. Gallagher, A. Ni Anniadh, K. Kruyere, M. Ottenio, H. Xie, and M. D. Gilchrist, “Dynamic tensile properties of human skin,” 2012 IRCOBI Conf., pp. 494–502, 2012.
[19] M. D. RIDGE and V. WRIGHT, “A Bio-Engineering Study of the Mechanical Properties of Human Skin in Relation to Its Structure.,” Br. J. Dermatol., vol. 77, no. 12, pp. 639–649, 1965.
[20] W. L. E. Wong, et al, “Resolving the viscoelasticity and anisotropy dependence of the mechanical properties of skin from a porcine model,” Biomech. Model. Mechanobiol., 2016.
[21] S. Schick, et al, “Maximum tensile stress and strain of skin of the domestic pig—differences concerning pigs from organic and non-organic farming,” Int. J. Legal Med., 2019.
[22] NATO/PFP, “Risk Assessment of non-lethal Kinetic Energy projectiles,” STANREC 4744
[23] A. Oukara, N. Nsiampa, C. Robbe, and A. Papy, “Injury Risk Assessment of Non-Lethal Projectile Head Impacts,” Open Biomed. Eng. J., vol. 8, no. 1, pp. 75–83, 2014.