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Toward Durable Prosthetic Feet: Mechanical and Tribological Evaluation of Polyurethane Elastomers

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Abstract. Elastomeric materials play a crucial role in determining the durability, comfort, and service life of prosthetic feet. Although various elastomers have been explored for this application, systematic evaluation of polyurethane formulations that integrate mechanical and tribological performance remains limited. This study investigates two types of polyurethane elastomers (A70 and A80) with different polyol–isocyanate ratios to identify an optimal formulation for prosthetic foot applications. Mechanical properties were evaluated through tensile, tear, hardness, and density tests, while friction and wear behavior were assessed using a pin-on-disk tribometer. A commercial prosthetic foot elastomer available in the Indonesian market was used as a benchmark. The results show that A80-based formulations generally provide superior tensile and tear strength compared to A70 variants. Among all specimens, A80-3 exhibits the lowest wear volume and stable friction performance, while maintaining hardness comparable to the commercial product. Although its density remains higher than the benchmark, the improved wear resistance indicates potential for extended service life. These findings offer practical guidance for durability-oriented material selection in the development of cost-efficient prosthetic feet.

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1. Introduction

Amputation is a medical treatment to remove a body part because of several reasons, such as diseases and injuries. Lower-limb amputation is a common type of amputation [1]. In 2017, an estimated 57.7 million people worldwide were living with limb amputation due to traumatic causes, 35.3 million of them had lower limb amputation, both unilateral and bilateral traumatic [2]. Lower limb amputation will significantly affect the quality of life since it will reduce mobility [3], loss of productive capacity [4], and psychological problems [5]. One of the solutions for patients with lower limb amputation is lower limb prostheses, which generally consist of a socket, pylon, and prosthetic foot.

The prosthetic foot is an important part since it interacts with ground and must be replaced regularly due to wear on the surface. During walking, the prosthetic foot experiences repeated cyclic loading and sliding contact with the ground, which gradually leads to material degradation and surface wear. Generally, studying materials for every product is important to improve the quality and lifetime of the product [6]. The quality of prosthetic foot, especially elastomeric materials, determines the lifetime of the product. Several prosthetic foot products are available in the Indonesian market, both locally and imported products. However, the quality of the local product is inferior compared to the imported products, including the aesthetic, weight, and lifetime [7]. Improving the material performance of locally developed prosthetic feet is therefore important to enhance durability and reduce replacement frequency. The prosthetic foot will have a longer lifetime if the elastomeric materials have high wear resistance. On the other hand, the elastomeric materials should have high friction to avoid slips. Therefore, the elastomeric materials selection of prosthetic foot is of importance.

Some types of elastomeric materials have been studied to be used in prosthetic feet, such as natural rubber [8] and silicone [9]. Previous studies have shown that silicone that is reinforced with talc powder increase their tensile strength but do not significantly improve tear resistance [9]. Natural rubber shows good tensile strength, tear strength, and friction, but its density is two times larger compared to commercial products [7]. In addition, the repeatability of natural rubber is also a challenge because it is a natural material that cannot be fully controlled [10, 11]. Although these studies have provided insight into the mechanical performance of elastomeric materials for prosthetic feet, systematic evaluation of their tribological behavior—particularly friction and wear characteristics under controlled sliding conditions—remains limited in the literature.

In the present study, elastomeric materials based on polyurethane were investigated in the mechanical and tribological aspects. For mechanical properties aspects, the study was carried out for tensile strength, tear strength, hardness, and density. While the tribological aspects were carried out for wear and friction. The elastomeric material of a commercial prosthetic foot product available in the Indonesian market was used as a benchmark in the present study. Polyurethane elastomers were selected due to their tunable mechanical properties through variation of polyol–isocyanate ratios, which allow optimization of stiffness, strength, and durability. This study aims to investigate how variations in polyurethane formulation influence tensile strength, tear resistance, hardness, density, friction, and wear behavior, and to identify the formulation that provides the best compromise between durability and functional performance for prosthetic foot applications. The novelty of this work lies in the systematic benchmarking of polyurethane formulations against a commercial prosthetic foot elastomer using combined mechanical and tribological evaluation to support engineering-based material selection.

2. Methods

2.1. Materials and fabrication

The materials used in the present study were Polyurethane elastomer A70 and A80 obtained from SKPU. These polyurethanes consist of Polyol and Isocyanate. The type of Polyol used in the present study is Polypropylene Glycol (PPG) polyether. While the type of Isocyanate used in the present study is Diphenyl Methane-4,4'-diisocyanate. The A70 and A80 grades were selected to represent two different hardness levels commonly used in elastomeric engineering applications, allowing investigation of the influence of formulation stiffness on mechanical and tribological behavior.

Fabrication of the specimens was carried out by mixing the Polyol and Isocyanate for 4 minutes at room temperature. The mixing process was conducted using mechanical stirring to ensure homogeneity of the reactive components. The resulting mixture was poured into the prepared mold. The mold was cleaned and coated by using a release agent to ensure the specimens could be taken off easily. The mold and mixture were placed into a vacuum chamber to remove the air during the curing process. The curing process was performed at room temperature (approximately 25 ± 2 °C) under ambient laboratory conditions. The specimens were allowed to cure for 24 hours before demolding and were further conditioned at room temperature for at least 48 hours prior to testing to ensure stabilization of mechanical properties.

The composition of Polyol and Isocyanate was varied to investigate the optimum formulation for foot prosthesis application, see Table 1. The selected polyol–isocyanate ratios were determined based on preliminary trials and manufacturer recommendations to produce elastomers within the targeted hardness range suitable for prosthetic foot applications.

Table 1. Variation of the ratio between Polyol and Isocyanate.

Specimen Code	Type of Polyurethane	Polyol: Isocyanate
A70 - 1	A70	1.1 : 1
A70 - 2	A70	1.2 : 1
A70 - 3	A70	1.3 : 1
A80 - 1	A80	0.9 : 1
A80 - 2	A80	1.0 : 1
A80 - 3	A80	1.1 : 1

2.2. Experimental tests

The mechanical characterization of the specimens was conducted for tensile test, tear test, hardness test, and density test. The tensile and tear tests were carried out with a Universal Testing Machine Tinius Olsen H25K follows ISO 37:2015 and ISO 34-1:2015, respectively. A Bareis Durometer was used for the hardness tests based on ISO 7619-1:2010. Whilst the density of the specimens was identified following ISO 2781:2008. Several molds were made to produce specimens which fit with the geometry requirements for all experimental tests. All tests were repeated three times to ensure the validity of the results.

For each formulation, three independent specimens (n = 3) were tested, and the results are presented as mean values with standard deviation. Statistical analysis was performed using one-way ANOVA with a significance level of 0.05 to determine whether differences among formulations were statistically significant.

A rotary pin-on-disc tribometer from Ducom Instrument was used to investigate the tribological phenomena of the specimens, i.e. friction and wear. The experimental tests follow the procedure of ASTM G99. The schematic of the pin-on-disc tribometer is depicted in Figure 1. The sliding contact occurs between a rigid sphere pin and a polyurethane disc. Prior to testing, both pin and specimen surfaces were cleaned to remove contaminants.

The load of 5 N was used in the sliding contact experiments. The angular speed of the disc is 200 rpm, and the radius of the wear track is 30 mm, resulting in a linear speed of 0.63 m/s. The tests were conducted at ambient temperature. The tribometer tests were stopped at 8 minutes, resulting in the sliding distance reaching 302 m. At this sliding distance, the friction force of all specimens has reached a steady state phase. The selected load and sliding speed were chosen to represent moderate contact conditions relevant to elastomer–ground interaction during normal walking, while maintaining stable and repeatable laboratory conditions.

In this study, the friction force of each specimen is defined as the friction force at a steady state. The friction force is obtained from the average value of the friction force during a steady state. The wear mass of the specimens was determined by weighing the specimens before and after the tribometer test. Each tribological test was also repeated three times for every formulation to ensure repeatability of the measured friction and wear values.



Figure 1. The schematic of pin-on-disc tribometer.

3. Results and Discussion

3.1. Tensile tests

The gait cycle process during walking consists of several phases, such as heel strike, mid stance, terminal stance, and toe-off. At the terminal stance phase, the foot prosthesis will experience a bending moment which results in tensile stress on the lower part of the foot. The tensile stress of elastomer can initiate cracks on the elastomeric surface and may propagate to failure. Therefore, the tensile strength of the elastomer is an important parameter when selecting the type of elastomer. In this study, the specimens were subjected to tensile test using the Universal Testing Machine Tinius Olsen H25K following ISO 37:2015.

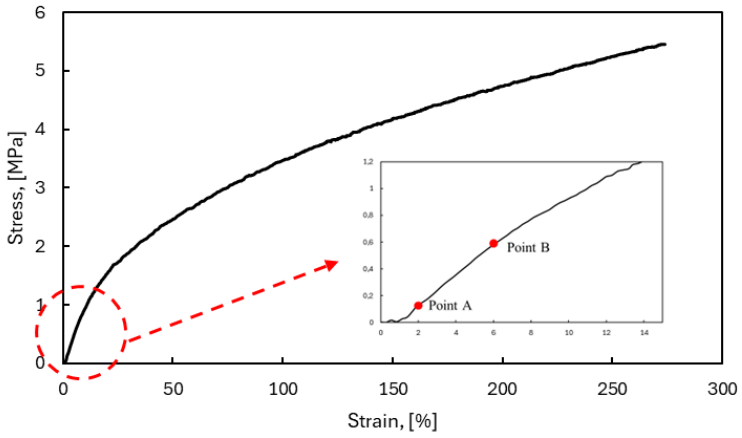


Figure 2. Stress-strain relation of A80-2 specimen.

The stress-strain relation of a tensile test result is shown in Figure 2. The fracture point is important because it shows the tensile strength and elongation of the elastomer when it experiences fracture. The

higher the tensile strength, the better the material to resist tensile failure. While the higher elongation at break will result in a better ability to strain before the elastomer experiences fracture.

Another important parameter in the stress-strain curve is the elastic modulus. Determining the elastic modulus of elastomeric materials is difficult due to its viscoelastic behavior, therefore the material actually does not really have an elastic phase. Several researchers identify the elastic modulus of elastomer at small strains, assuming the elastic region occurs at that strain [12]. In the present study, the elastic modulus of the specimens was calculated at a small strain, see Figure 2.

Figure 3 shows the average tensile strength of three tests for every specimen. In general, the tensile strength of the A80 polyurethane is higher than that of the A70 polyurethane. Variations in the amount of polyol show the same results for both types of materials, the higher amount of polyol will reduce the tensile strength. Polyol is flexible segments of the polymeric chain that contribute to the elasticity of polyurethanes. If polyol is added in excess, the final polyurethanes tend to be softer due to reduced crosslink density [13]. Quantitatively, the tensile strength of A80-based formulations is higher than that of the commercial elastomer benchmark, indicating improved resistance to tensile-induced crack propagation. Among all specimens, A80-1 exhibits the highest tensile strength (8 ± 0.34 MPa), which is approximately higher than the benchmark material. One-way ANOVA analysis reveals a statistically significant difference in tensile strength among polyurethane formulations ($F = 15.58$, $p < 0.001$), confirming that variation in polyol–isocyanate ratio plays a decisive role in determining mechanical performance.

In the standard of ISO 10328 about structural testing of lower-limb prostheses, there is no specific requirement about the mechanical properties of elastomer used in the prosthetic foot [14]. Therefore, the benchmark in the present study used the mechanical properties of an elastomer of a commercial prosthetic foot product that is available in the Indonesian market. The dashed line in Figure 3 represents the tensile strength of the elastomer used by the commercial product. From an engineering perspective, higher tensile strength contributes to improved durability under cyclic bending loads experienced during walking, potentially reducing the risk of premature material failure.

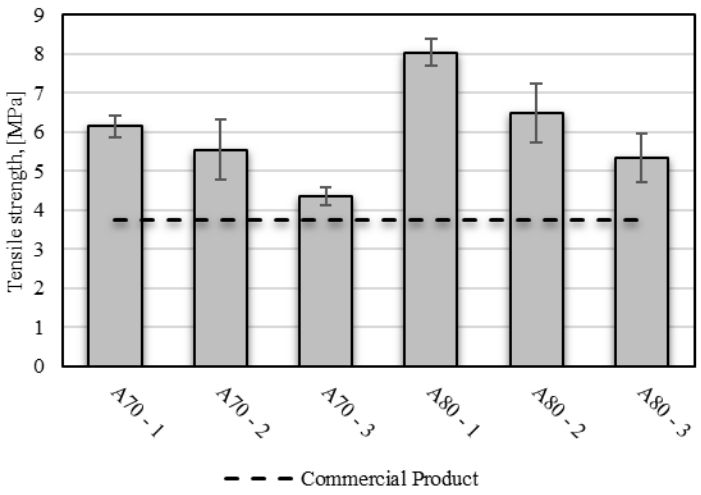


Figure 3. The tensile strength of all specimens compared to the commercial product.

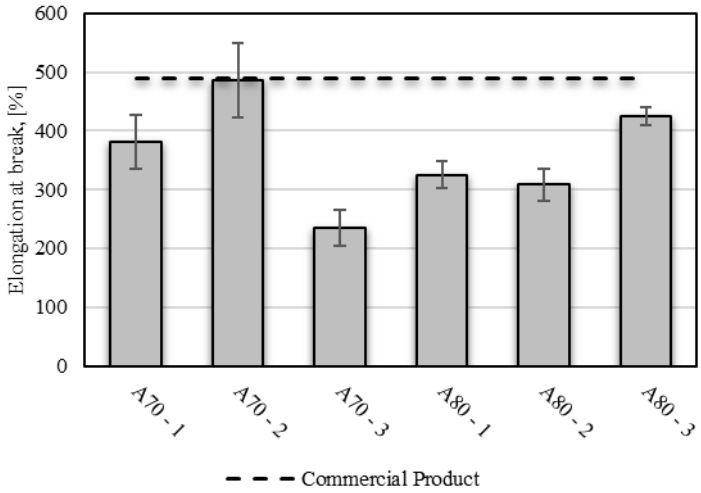


Figure 4. The elongation at break of all specimens compared to the commercial product.

The variation of the ratio of polyol and isocyanate does not show a particular trend for the elongation at break, see Figure 4. All specimens exhibited elongation at break values exceeding 200%, indicating sufficient ductility for elastomeric applications. The highest elongation was obtained in A70-2, while the lowest was observed in A70-3.

The elongation at break of A70-2 is comparable to that of the commercial elastomer, suggesting similar deformability under tensile loading. The absence of a clear trend may indicate that elongation at break is influenced not only by polyol content but also by microstructural factors such as phase separation and crosslink density distribution within the polyurethane network.

Although lower elongation at break may increase the risk of failure under extreme deformation, the typical strain experienced by a prosthetic foot during walking is considerably lower than the measured fracture strain [15]. Therefore, all tested formulations provide sufficient deformability for practical prosthetic foot applications, and tensile strength and wear resistance become more dominant selection criteria.

3.2. Hardness Tests

The hardness test results for all specimens are shown in Figure 5. In general, the hardness of the A-80 polyurethane is higher than that of the A-70 polyurethane. Figure 5 also shows that the higher the polyol, the lower the hardness. This behavior can be attributed to the role of polyol as the soft segment in polyurethane chains, whereas isocyanate contributes to the formation of hard segments through crosslinking reactions [13].

From a materials engineering perspective, higher hardness generally correlates with improved resistance to surface deformation under contact loading. Quantitatively, all tested formulations exhibit higher hardness than the commercial elastomer benchmark, indicating increased stiffness of the polyurethane network. However, excessively high hardness may reduce compliance and potentially affect comfort during walking due to reduced energy absorption capacity. Among all specimens, A70-3 shows hardness values closest to the commercial product, suggesting a comparable balance between stiffness and flexibility.

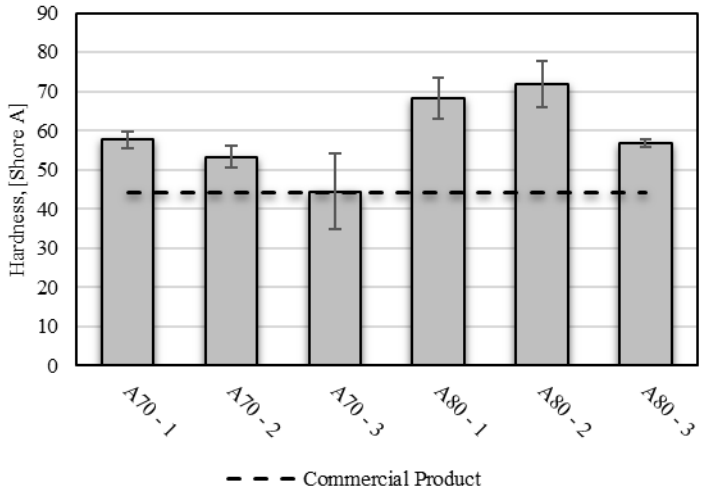


Figure 5. The hardness of all specimens compared to the commercial product.

3.3. Tear Tests

The elastomeric material used in prosthetic foot products must have good tear strength because it may come into contact with sharp objects during daily use. Sharp objects may initiate cracks on the elastomer surface and may propagate into tears if there is a tensile load exerted on the material. Tear testing quantifies the force required to propagate a tear per unit thickness (N/mm), and therefore provides an important indicator of crack-growth resistance in elastomeric components. From an engineering perspective, higher tear strength is particularly relevant for prosthetic feet because localized damage at edges or tread features can develop into larger defects under repeated walking cycles.

In previous research [9], material optimization using several silicone elastomers reinforced with talc powder showed that the mechanical properties of the material such as hardness and tensile strength increased, but the tear strength of the material did not show a significant increase, approximately 1.5 N/mm.

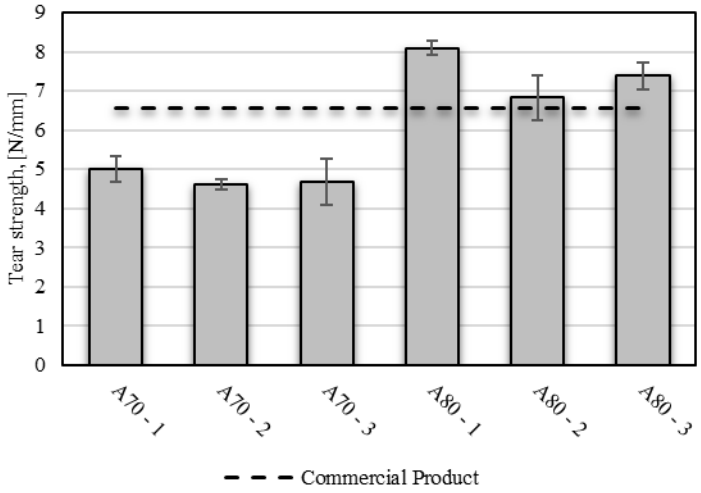


Figure 6. The tear strength of all specimens compared to the commercial product.

Figure 6 shows that the tear strength of the A-80 polyurethane is higher (approximately 7 N/mm) compared to the A-70 polyurethane (approximately 5 N/mm). Across the investigated polyol–isocyanate ratios, the tear strength shows no clear monotonic trend, suggesting that tear resistance is influenced not

only by formulation ratio but also by microstructural factors such as phase separation and hard–soft segment distribution. Only the tear strength of A80 polyurethane exceeds the tear strength of the commercial product. While the tear strength of A70 polyurethane is below the commercial product. These results also indicate that all polyurethane specimens have a higher tear strength compared to silicone material in previous studies, 1,5 N/mm [9]. Overall, the relatively high tear strength of A80 formulations supports their suitability for durable prosthetic foot applications where resistance to crack propagation is required.

3.4. Density

One of the requirements of a prosthetic foot product is low weight to improve walking comfort and reduce metabolic energy expenditure [16], therefore low density of material is needed. Excessive mass in the distal part of the prosthesis increases swing effort during gait and may negatively affect user comfort and mobility. Some previous studies revealed that the design modification and material combination are also visible to decrease the weight of the product [17, 18].

The density of the specimens was measured using a digital balance with an accuracy of 0.001 grams, following the ASTM D792 standard. While the volume of the specimen was measured by using the principle of Archimedes' Law. Each measurement was repeated three times, and the reported values represent the average density of the specimens. The density test aims to determine the mass per unit volume of a particular specimen.

The results showed that the density of all specimens was higher than that of commercial products (0.54 gr/cm^3). The increase in density with higher polyol content may be associated with variations in molecular packing and network structure within the cured polyurethane. The specimen that has the closest density to the commercial product is the A80-1, see Figure 7. Although A80-based formulations demonstrate superior mechanical and wear performance, their higher density represents a design trade-off that must be addressed to maintain user comfort. To mitigate this limitation, lightweight structural configurations such as hollow cores, graded materials, or hybrid material combinations may be employed to reduce overall prosthetic foot mass without compromising surface durability [19].

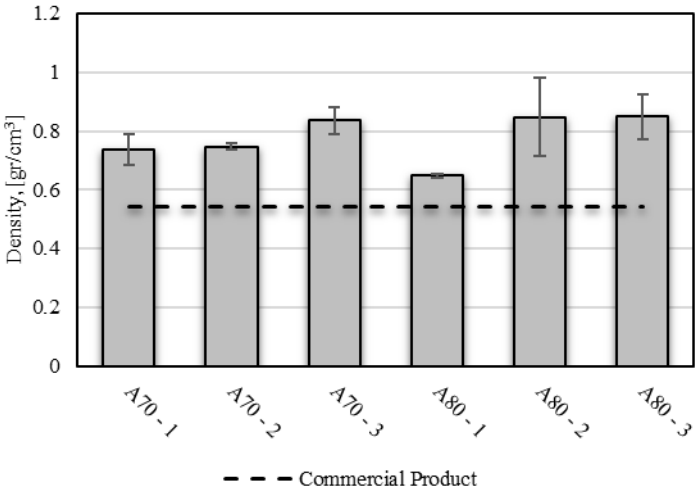


Figure 7. The density of all specimens compared to the commercial product.

3.5. Tribometer Tests

During walking, the prosthetic foot is continuously subjected to contact and sliding interaction with the ground surface. A good product requires sufficient friction so that the prosthetic foot will not be slippery during usage. At the same time, a prosthetic foot also requires good wear resistance to extend the lifetime of the product. These two requirements are often interrelated, as higher friction may increase surface stresses and potentially accelerate material degradation. In this study, a pin-on-disk tribometer was

employed to systematically evaluate the friction and wear behavior of the investigated polyurethane formulations under controlled laboratory conditions.

Figure 8 shows a graph of the friction force as a function of the sliding distance. The total friction force consists of two contributors, i.e. adhesive friction (F_{adh}) and deformation friction (F_{def}) [20]. The adhesive friction depends on the real contact area (A_r) and shear stress (τ), $F = A \cdot \tau$. While the deformation friction depends on the energy dissipation which correlate to the elastic modulus of the materials [21]. Depending on the operating conditions, both types of friction forces can play a role in elastomer contact [22].

At the beginning of the test, the friction force increases. This phenomenon can be caused by a lot of wear occurring on the polyurethane surface and as a result, the real contact area between the two contacting materials increases. The larger the real contact area, the larger the friction force. After a certain sliding distance, the wear will decrease gradually because the polyurethane surface becomes smoother due to the wear of asperities. The friction force will reach its peak and experience a slight decrease before reaching the steady state phase. The slight decrease in friction force can be caused by several reasons, such as debris between the contacting materials that act like rolls [23] and the formation of a solid lubricant layer on the material [12, 24]. In this study, the steady-state friction force is defined as the average friction force during this stable region.

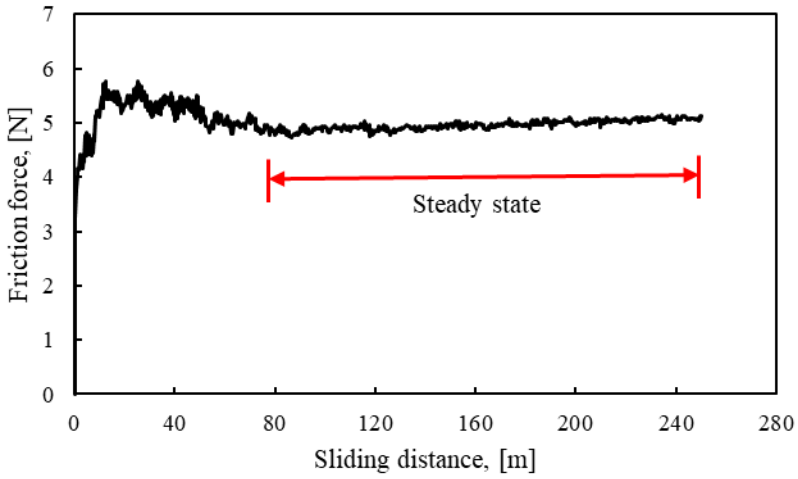


Figure 8. Friction force as a function of sliding distance for A70-3 polyurethane.

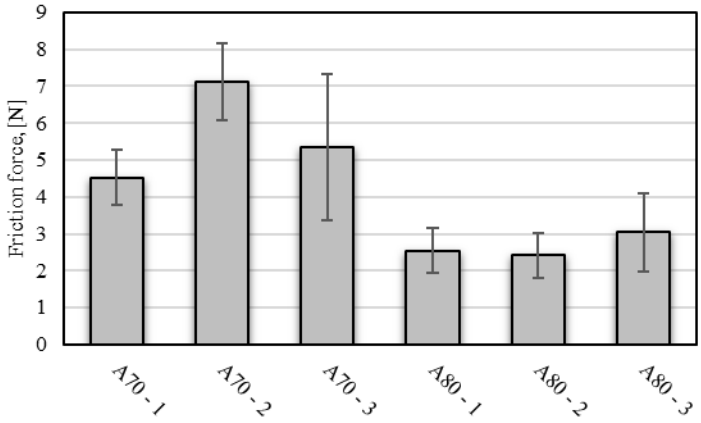


Figure 9. The friction force of all specimens.

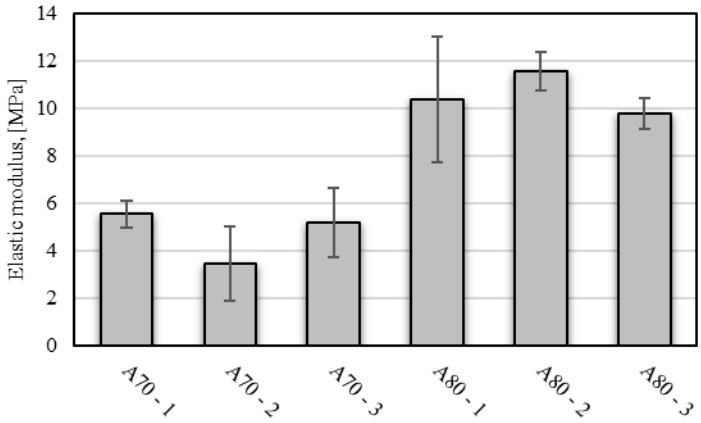


Figure 10. The elastic modulus of all specimens.

For elastomeric materials used in prosthetic feet, sufficient friction is required to prevent slipping during walking; however, excessively high friction may accelerate wear and increase energy dissipation. Therefore, an optimal elastomer formulation should provide balanced friction behavior to ensure safety without compromising durability. The friction force at the steady state phase of all specimens is shown in Figure 9. In general, the friction force of the A70 polyurethane has a higher value compared to the A80 polyurethane. This behavior is consistent with the lower elastic modulus (Figure 10) and hardness (Figure 5) of A70 formulations. One-way ANOVA analysis indicates that the steady-state friction force differs significantly among polyurethane formulations ($F = 8.25$, $p < 0.01$), confirming that formulation variation significantly affects tribological behavior.

Lower hardness is generally associated with higher friction forces in elastomer–solid contacts [25], as softer materials tend to deform more easily under normal load. For elastomers with lower elastic modulus, the real contact area increases under the same normal load, thereby enhancing adhesive friction due to greater interfacial shear interaction [26]. In addition to adhesive friction, deformation friction also contributes significantly to the total friction force in elastomeric materials. A lower modulus (A70 polyurethane) undergoes larger viscoelastic deformation during sliding contact. This increased deformation leads to greater internal energy dissipation, which further raises the friction force. Therefore, the combined effect of enhanced adhesive friction and increased deformation friction explains why A70 polyurethane exhibits higher steady-state friction compared to A80 polyurethane.

The elastic modulus also affects the comfort of the prosthetic foot product. A higher elastic modulus results in a stiffer material which may limit dorsiflexion and plantarflexion movements during gait. Conversely, a lower modulus improves compliance but may increase friction and wear. Thus, material selection for prosthetic foot applications requires careful balancing between friction performance, mechanical stiffness, and functional comfort.

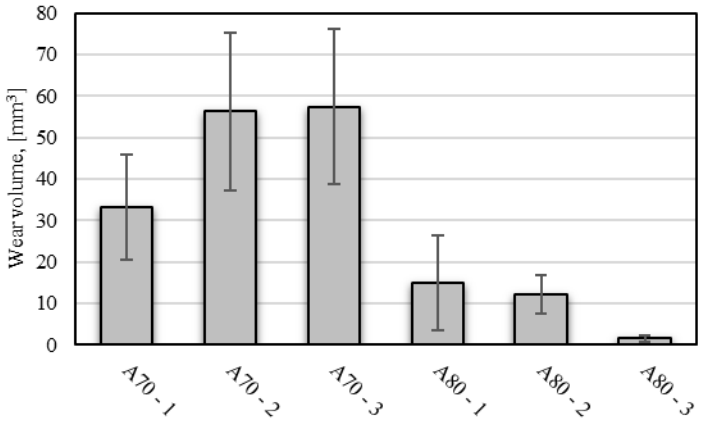


Figure 11. The wear volume of all specimens.

The wear mass of the specimen is obtained by weighing the specimen before and after the tribological testing. The wear volume is obtained by calculating the wear mass and density of each specimen. The density of each specimen used in the calculation is shown in Figure 7. The results of the wear volume for all specimens are shown in Figure 11, it can be observed that the wear volume of A70 polyurethane is much higher than that of A80 polyurethane. The wear volume results are inversely proportional to the hardness and elastic modulus results, see Figure 5 and Figure 10. In general, materials with higher hardness and stiffness are less prone to surface damage under sliding contact.

One-way ANOVA analysis confirms that wear volume differs significantly among polyurethane formulations ($F = 8.52$, $p < 0.01$), indicating that variation in polyol–isocyanate ratio plays a decisive role in determining wear resistance. This statistical result supports the observed trend that A80 formulations provide superior tribological durability compared to A70 formulations. This finding is in agreement with previous studies on polyurethane elastomers, where increased stiffness contributes to improved wear resistance [27].

From a tribological standpoint, harder materials tend to have better wear resistance because they will reduce abrasive wear [28]. In contrast, softer materials such as A70 polyurethane experience larger deformation, which may facilitate micro-cutting or fatigue-related surface damage over repeated cycles. As a result, material loss becomes more significant.

For prosthetic foot applications, enhanced wear resistance is important because the component is repeatedly exposed to ground contact during walking. Lower wear volume may help maintain surface integrity for a longer period and potentially reduce the frequency of replacement. Among all tested formulations, A80-3 exhibits the lowest wear volume, suggesting that this formulation provides the most promising durability performance under the present test conditions.

3.6. Material selections

The performance of prosthetic foot products is regulated by the ISO 10328 standard, which specifies structural tests such as static and fatigue testing [14]. However, the standard does not define specific mechanical or tribological requirements for elastomeric materials used in prosthetic feet. Therefore, material selection must be based on a balanced evaluation of mechanical strength, tribological behavior, and functional considerations. For prosthetic foot applications, important properties required by elastomer materials are tensile strength, tear strength, hardness, density, friction, and wear. Tensile strength, tear strength, friction, and wear are related to product durability, whereas hardness and density influence stiffness, comfort, and overall component weight.

Statistical analysis confirms that formulation variation significantly affects both mechanical and tribological performance. One-way ANOVA results indicate significant differences among formulations for tensile strength ($p < 0.001$), steady-state friction force ($p < 0.01$), and wear volume ($p < 0.01$). These findings demonstrate that the polyol–isocyanate ratio plays a critical role in governing overall elastomer

performance.

Based on the experimental results, all specimens exhibit higher tensile strength (see Figure 3) and hardness (see Figure 5) compared to commercial products. However, density remains the main limitation, as all polyurethane formulations show higher values than the commercial elastomer, although A80-1 approaches the benchmark value. In practical application, the overall mass of the product can be reduced by combining several materials that have a lower density in the inner structure of the prosthetic foot [19].

Considering that prosthetic feet are subjected to repeated ground contact and surface degradation during daily use, wear resistance becomes a critical parameter for extending service life and reducing replacement frequency, followed by tensile and tear strength. Based on the results of this study, the A80-3 polyurethane is the most optimum candidate because it has the best wear resistance (Figure 11), while maintaining tensile and tear strength superior to the commercial product. In addition, A80-3 polyurethane exhibits hardness values comparable to the benchmark and friction levels sufficient for safe ground interaction. Although its density is approximately 1.5 times higher than that of the commercial elastomer, the superior wear resistance and mechanical performance indicate that A80-3 provides the most balanced combination of durability and functional properties among the investigated materials. Therefore, A80-3 polyurethane is identified as the most suitable candidate under the present evaluation criteria, while further structural optimization is recommended to address weight considerations.

4. Conclusion

This study evaluated the mechanical and tribological properties of polyurethane elastomers intended for prosthetic foot applications and compared them with a commercial product. The results indicate that variations in the polyol–isocyanate ratio influence tensile strength, hardness, friction, and wear behavior, confirming that formulation plays an important role in determining overall elastomer performance.

Among the tested formulations, A80-3 demonstrates the lowest wear volume while maintaining adequate tensile and tear strength, as well as friction levels suitable for ground interaction. From a durability perspective, this formulation provides a reasonable balance between mechanical strength and tribological resistance. However, its higher density compared to the commercial elastomer remains a limitation that may require structural or design adjustments.

The present study is limited to laboratory-scale testing under a single loading condition. Further work involving fatigue testing, environmental exposure, and evaluation on full prosthetic foot prototypes would be beneficial to better understand long-term performance in real use conditions.

Declaration of AI and AI assisted technologies in the writing process

During the preparation of this work the author(s) used ChatGPT in order to improve the clarity, grammar, and readability of the manuscript. After using this tool/service, the author(s) reviewed and edited the content as needed and take(s) full responsibility for the content of the publication

Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper

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References

[1] Selvam PS, Sandhiya M, Chandrasekaran K, Rubella DH, Karthikeyan S. Prosthetics for lower limb amputation. Prosthetics and Orthotics: IntechOpen; 2021. https://doi.org/10.5772/intechopen.95593

[2] McDonald CL, Westcott-McCoy S, Weaver MR, Haagsma J, Kartin DJP, international o. Global prevalence of traumatic non-fatal limb amputation. 2021;45(2):105-14. https://doi.org/10.1177/0309364620972258

[3] Morgan SJ, Liljenquist KS, Kajlich A, Gailey RS, Amtmann D, Hafner BJ. Mobility with a lower limb prosthesis: experiences of users with high levels of functional ability. Disability and Rehabilitation. 2022;44(13):3236-44. https://doi.org/10.1080/09638288.2020.1851400

[4] Silva AdM, Furtado G, Dos Santos IP, da Silva CB, Caldas LR, Bernardes KO, et al. Functional capacity of elderly with lower-limb amputation after prosthesis rehabilitation: a longitudinal study. 2021;16(5):556-60. https://doi.org/10.1080/17483107.2019.1684581

[5] Şimsek N, Öztürk GK, Nahya ZN. The Mental Health of Individuals With Post-Traumatic Lower Limb Amputation: A Qualitative Study. 2020;7(6):1665-70. https://doi.org/10.1177/2374373520932451

[6] Mohammed R, Shaik AS, Mohammed S, Bunga KK, Aggala C, Babu BP, et al. Advancements and Challenges in Additive Manufacturing: Future Directions and Implications for Sustainable Engineering. Advance Sustainable Science, Engineering and Technology. 2025;7(1). https://doi.org/10.26877/asset.v7i1.1079

[7] Khafidh M, Marwa PA, Alfajr B, Suryawan D, Kistriyani L, Ismail R. The mechanical properties of elastomeric materials in prosthetic feet: Comparison between local and imported products. AIP Conference Proceedings. 2023;2568(1). https://doi.org/10.1063/5.0114404

[8] Kistriyani L, Khafidh M, Suryawan D, Ismail R, editors. Physical characteristic of polymer formulations for prosthetic foot materials: Comparison of natural rubber and ethylene vinyl acetate. Proceeding of The 2nd International Conference on Engineering Science and Technology (ICEST 2021); 2023: AIP Publishing LLC. https://doi.org/10.1063/5.0109385

[9] Khafidh M, Suryawan D, Kistriyani L, Naufal M, Ismail R. Friction Optimization of Talc Powder-Reinforced Elastomers for Prosthetic Foot Application. 2023;5(1):88-99. https://doi.org/10.3390/prosthesis5010007

[10] Omar MF, Ali F, Jami MS, Azmi AS, Ahmad F, Marzuki MZ, et al. A Comprehensive Review of Natural Rubber Composites: Properties, Compounding Aspects, and Renewable Practices with Natural Fibre Reinforcement. 2025;13(3). https://doi.org/10.32604/jrm.2024.057248

[11] Guerra NB, Sant'Ana Pegorin G, Boratto MH, de Barros NR, de Oliveira Graeff CF, Herculano RD. Biomedical applications of natural rubber latex from the rubber tree Hevea brasiliensis. Materials Science and Engineering: C. 2021;126:112126. https://doi.org/10.1016/j.msec.2021.112126

[12] Khafidh M, Schipper DJ, Masen MA. The Formation of a Modified Surface Layer on Elastomeric Materials. Tribology Letters. 2019;67(1):27. https://doi.org/10.1007/s11249-019-1140-4

[13] de Souza FM, Kahol PK, Gupta RK. Introduction to Polyurethane Chemistry. Polyurethane Chemistry: Renewable Polyols and Isocyanates. ACS Symposium Series. 1380: American Chemical Society; 2021. p. 1-24. https://doi.org/10.1021/bk-2021-1380.ch001

[14] ISO 10328: 2016 Prosthetics. Structural testing of lower-limb prostheses. Requirements and test methods. London, UK.

[15] Subhash N, Mehra N, Choudhary A, Baby CJ, Aravind A, Benny P, et al. Structural Performance Assessment of FUPRO Grace Foot. 2022;36(2).

[16] Kathrotiya D, Yusuf A, Bhagchandani RK, Gupta S. A Study for the development of prosthetic foot by additive manufacturing. Journal of the Brazilian Society of Mechanical Sciences and Engineering. 2023;45(3):187. https://doi.org/10.1007/s40430-023-04107-y

[17] Ismail R, Nursafitri MA, Fardinansyah AP, Fitriyana DF, Bayuseno AP, Siregar JP, et al. Design, fabrication, and performance testing of an energy storage and return (ESAR) foot prosthesis made of prepreg carbon composite. 2025;5(1):20-32. https://doi.org/10.31603/mesi.12652
[18] Kareem ZJJoUoBfES. From Metals to Composite Materials: A Comprehensive Review of Materials Used in Prosthetic Limb Manufacturing. 2025;33(4):140-67. https://doi.org/10.29196/jubes.v33i4.5902
[19] Yousif LE, Abed MS, Al-Zubidi AB, Resan KK. Innovations in prosthetic foot design enhancing durability, functionality and comfort through PLA composite filament 3D printing. Pigment & Resin Technology. 2024;ahead-of-print(ahead-of-print). https://doi.org/10.1108/PRT-10-2023-0092
[20] Kaiser F, Savio D, Bactavatchalou RJL. Modelling of static and dynamic elastomer friction in dry conditions. 2024;12(7):250. https://doi.org/10.3390/lubricants12070250
[21] Markov DPJF. Adhesion at friction and wear. 2022;10(11):1859-78. https://doi.org/10.1007/s40544-021-0564-7
[22] Lyashenko IA, Pham TH, Popov VLJL. Transition between friction modes in adhesive contacts of a hard indenter and a soft elastomer: an experiment. 2024;12(4):110. https://doi.org/10.3390/lubricants12040110
[23] Khafidh M, Schipper DJ, Masen MA, Vleugels N, Dierkes WK, Noordermeer JWM. Friction and wear mechanism of short-cut aramid fiber and silica reinforced elastomers. Wear. 2019;428-429:481-7. https://doi.org/10.1016/j.wear.2019.04.016
[24] Kumar R, Hussainova I, Rahmani R, Antonov MJM. Solid lubrication at high-temperatures—A review. 2022;15(5):1695. https://doi.org/10.3390/ma15051695
[25] Pieniak D, Jedut R, Gil L, Kupicz W, Borucka A, Selech J, et al. Comparative Evaluation of the Tribological Properties of Polymer Materials with Similar Shore Hardness Working in Metal–Polymer Friction Systems. 2023;16(2):573. https://doi.org/10.3390/ma16020573
[26] Maksuta D, Dalvi S, Gujrati A, Pastewka L, Jacobs TD, Dhinojwala AJSM. Dependence of adhesive friction on surface roughness and elastic modulus. 2022;18(31):5843-9. https://doi.org/10.1039/d2sm00163b
[27] Junik K, Lesiuk G, Barcikowski M, Błażejewski W, Niemiec A, Grobelny M, et al. Impact of the hardness on the selected mechanical properties of rigid polyurethane elastomers commonly used in suspension systems. Engineering Failure Analysis. 2021;121:105201. https://doi.org/10.1016/j.engfailanal.2020.105201
[28] Ligier K, Olejniczak K, Napiórkowski J. Wear of polyethylene and polyurethane elastomers used for components working in natural abrasive environments. Polymer Testing. 2021;100:107247. https://doi.org/10.1016/j.polymertesting.2021.107247