Human Lander Challenge: Microgravity Blood Infusion Pump

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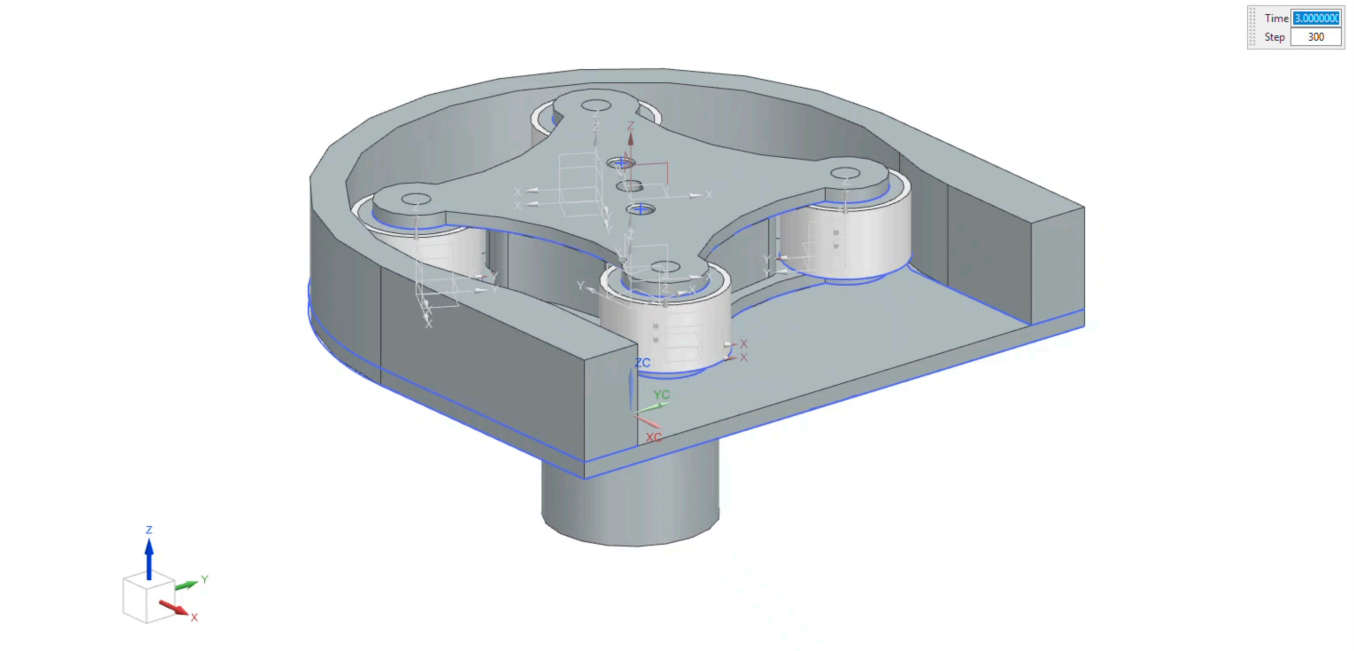

Purdue University - Microgravity Blood Infusion Pump

Microgravity Medical Systems → Blood Transfusion Capability for Long-Duration Spaceflight

Major Objectives:

- Enable transfusion in microgravity/partial gravity
- Prevent air embolism + contamination; limit hemolysis
- Deliver 400–500 mL at 100–250 mL/hr with real-time monitoring

Technical Approach:

- Peristaltic pump (gravity-independent, sterile tubing path)
- PTFE membrane bubble trap + vacuum assist; inline bubble sensor w/ auto shutoff
- Arduino control + feedback sensors (flow/pressure/air/battery)

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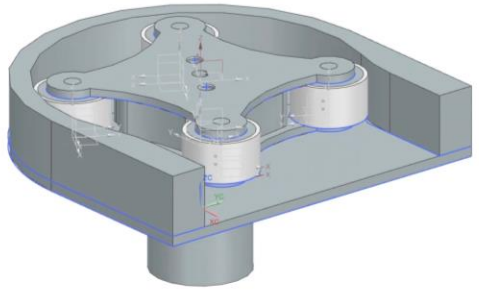

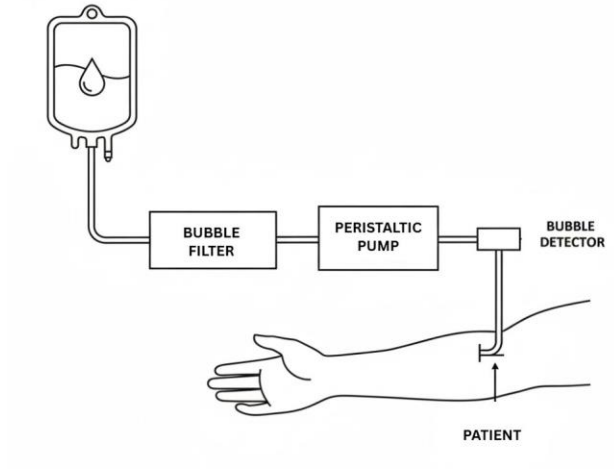

Key Design Details:

- Peristaltic pump (sterile tubing path) for controlled microgravity flow
- PTFE membrane bubble trap + vacuum assist
- Inline bubble sensor with auto shutoff/alarm + Arduino control

Innovations:

- Gravity-independent air removal via membrane separation, with the required pressure differential created by the vacuum pump.
- Built-in safety redundancy: remove bubbles and detect/stop remaining air
- Modular compact packaging with a total volume of 512 in^3, designed around spacecraft mass/volume constraints while supporting emergency transfusion.

Schedule (Path to Adoption):

- Years 0–1: lab prototype + subsystem tests (flow, bubble removal/detection, power) → TRL 3→4
- Years 1–3: vibration/thermal vacuum + reduced-gravity testing → TRL 4→5
- Years 3–5: flight-representative engineering model + integrated testing → TRL 5→6
- Years 5–8: qualification + spacecraft integration → TRL 7–8

Costs (Prototype):

- Hardware subtotal: ~$1,169
- Testing/contingency (~20%): ~$234
- Total prototype cost: ~$1,403

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1 EXCUTIVE SUMMARY

As human exploration missions extend beyond low Earth orbit under NASA's Artemis campaign, medical autonomy becomes essential for crew survival. Blood transfusion capability is critical for treating hemorrhage and severe anemia during long-duration lunar and Mars missions, yet conventional gravity-dependent infusion systems pose significant safety risks in microgravity due to air embolism and unstable fluid behavior.

This project presents a gravity-independent blood infusion pump designed to safely deliver rehydrated, freeze-dried blood products in space. The system integrates a peristaltic pump for precise flow control with a microporous PTFE membrane air trap and ultrasonic bubble detection to actively mitigate embolism risk. The compact, closed-loop architecture operates within spacecraft constraints and supports autonomous medical operations.

With a structured verification plan and TRL advancement roadmap, this design provides a scalable, flight-viable solution that enhances crew safety, mission resilience, and long-duration exploration capability.

2 BACKGROUND AND ARTIMIS ALIGNMENT

2.1 CURRENT STATE OF THE ART

As human exploration extends beyond low Earth orbit, medical autonomy becomes increasingly critical. During sustained lunar missions and future Mars transit operations, evacuation to Earth will be difficult to impossible. Crews must therefore be equipped to manage traumatic injury, internal bleeding, and space-induced physiological conditions independently.

Microgravity introduces unique challenges to blood transfusions. Terrestrial transfusion systems rely on gravity-assisted flow and buoyancy-driven air separation within intravenous lines. In microgravity, however, buoyancy forces are negligible, and fluid behavior becomes dominated by surface tension. Gas bubbles do not naturally rise and separate from liquid blood, increasing the risk of air embolism during infusion. This fundamentally alters the safety profile of conventional transfusion hardware.

Long-duration missions require compact and stable blood storage strategies. Freeze-dried (lyophilized) blood products have been identified as a promising solution due to their extended shelf life and reduced storage mass compared to refrigerated blood. Lyophilization is likely to be turbulent, and the rehydration process is likely to introduce air bubbles into the blood, past studies have demonstrated visible bubbles in the reconstitution of lyophilized biological products (Gao 2025). Importantly, parabolic flight research has demonstrated that rehydration of freeze-dried blood products in microgravity is feasible (Elder 2024), validating the practicality of long-term blood storage for space missions. However, while

rehydration has been shown to be achievable, the safe and controlled delivery of reconstituted blood in microgravity remains insufficiently characterized.

2.2 ALIGNMENT WITH NASA EXPLORATION GOALS

NASA's Artemis campaign aims to establish a sustained human presence on the lunar surface while preparing for eventual Mars missions. These missions introduce operational durations ranging from approximately 30 days on the Moon to 1,200 days for Mars transit and surface operations. During these missions, crew survivability depends on enhanced medical autonomy and reduced reliance on Earth-based intervention.

A hemorrhagic event or severe anemia case during lunar or Mars missions could rapidly escalate to a life-threatening scenario without transfusion capability. Communication delays during Mars transit further limit real-time medical guidance from Earth, reinforcing the need for autonomous, reliable medical systems.

The proposed gravity-independent blood transfusion system directly supports Artemis objectives by:

- Increasing crew survival capability during medical emergencies
- Enabling long-duration mission autonomy
- Integrating with freeze-dried blood logistics to reduce resupply mass
- Operating across microgravity transit and reduced-gravity lunar or Martian surface environments
- Maintaining compatibility with spacecraft mass, volume, and power constraints

By addressing a currently unfilled gap in space medical capability, this system enhances mission resilience and supports NASA's long-term strategy for sustainable human exploration architectures.

3 NEEDS AND REQUIREMENTS

3.1 STAKEHOLDERS-DRIVE MISSION OBJECTIVES

System requirements were developed using a structured top-down systems engineering process beginning with stakeholder identification. Primary stakeholders include crew medical operators, flight surgeons, habitat integration engineers, mission planners, and safety authorities.

From stakeholder analysis, five primary mission objectives were defined:

1. Enable safe blood transfusion across microgravity, partial gravity, and terrestrial environments.
2. Prevent air embolism, contamination, and hemolysis during infusion.

3. Deliver clinically controlled transfusion volumes and flow rates consistent with medical protocols.
4. Provide mission-level reliability and reusability for repeated transfusion cycles.
5. Integrate within spacecraft mass, volume, power, and operational constraints to support rapid emergency deployment.

These objectives were translated into quantifiable platform-level requirements to ensure traceability and verifiability.

3.2 DERIVED PLATFORM REQUIREMENTS

Platform Functional Requirements (PFRs) were derived from the Primary Mission Objectives (PMOs) through structured functional decomposition. Each requirement is written to be measurable, testable, and traceable to stakeholder needs. Verification methods include analysis, inspection, demonstration, and testing to ensure compliance with both medical and spaceflight constraints.

All patient-contacting materials comply with ISO 10993 biocompatibility standards, and system risk management follows ISO 14971 guidance. Launch survivability requirements are verified through vibration and shock qualifications per applicable launch provider specifications.

Category	Requirement ID	Requirement Statement	Verification Method
Gravity Performance	PFR_5	The system shall support transfusion in microgravity (0 g).	Environmental testing
Gravity Performance	PFR_6	The system shall support transfusion in partial gravity (<1 g).	Environmental testing
Clinical Safety	PFR_14	The system shall detect and automatically stop flow for air volumes ≥ 20 μL.	Bubble challenge test
Clinical Safety	PFR_16	The system shall limit hemolysis to ≤ 0.8% during transfusion.	Laboratory testing
Transfusion Performance	PFR_23	The system shall deliver 400–500 mL per transfusion.	Flow capacity test
Transfusion Performance	PFR_24	The system shall maintain infusion rates between 100–250 mL/hr.	Flow validation
Reliability	PFR_9	The system shall support ≥ 50 transfusion cycles without critical mechanical or electrical failure.	Lifecycle testing
Integration Constraints	PFR_11	The system mass shall not exceed 10 lb.	Mass measurement
Integration Constraints	PFR_12	The system shall maintain a stowed volume not exceeding 512 in^3	Volume analysis
Monitoring & Autonomy	PFR_18	The system shall provide real-time monitoring of flow rate, pressure, air-in-line, and battery status.	Functional verification

4 System Design and Architecture

4.1 Pump Selection

Precise, gravity-independent control of rehydrated red blood cell (RBC) infusion (100–250 mL/hr) requires an active pumping mechanism. Passive drip systems are unsuitable in microgravity due to the absence of hydrostatic pressure. Four pump architectures were evaluated: gear, diaphragm, syringe, and peristaltic.

Peristaltic pumping was selected due to:

- Closed sterile fluid path (fluid contacts only disposable tubing)
- Gravity-independent positive displacement
- Low contamination risk
- Compatibility with reusable hardware

Gear pumps are considered unsuitable for pumping blood primarily because they induce high shear stress, which may lead to significant hemolysis. Diaphragm pumps introduce fluid–mechanical contact, increasing contamination risk. Syringe pumps offer high precision but are limited in volume capacity and are less suitable for repeated multi-unit transfusions. Past flight tests have demonstrated success with peristaltic pumps in moving intravenous fluid (Spaulding 2004).

For our flow range of 100-250 ml/hr we need a rotational speed of 4-10 RPM (see Appendix A for calculations).

We do acknowledge that there is some risk with the compression forces of a peristaltic pump; however, low rotational speeds and compliant medical tubing minimize shear forces, supporting hemolysis limits provided by the Council of Europe guideline, which can be verified through external testing.

4.2 AIR BUBBLE TRAP

A major challenge of performing blood transfusions in space is preventing air embolisms caused by microgravity fluid behavior. Initial concepts explored removing bubbles by configuring the tubing with sharp corners and internal vanes to force phase separation. However, this method was rejected because the necessary geometry would make the system too large and heavy for a portable medical device. Another alternative involved using a fine mesh to break larger air bubbles down into smaller, sub-clinical sizes. This approach was also discounted because, without gravity and buoyancy forces, the smaller bubbles remain suspended in the fluid path. This increases particle collisions and allows the microbubbles to recombine into hazardous volumes before reaching the patient. Clinical safety thresholds require bubbles to stay well under a volume of 20 μL (~3.4 mm in diameter) to pass through the lungs safely, though risks are tracked down to 0.004 μL (~0.2 mm) (Wilkins 2012). Because mechanical fragmentation cannot guarantee that bubbles will not recombine, a mesh does not fully ensure patient safety.

To overcome these limitations, the chosen design implements an inline air trap utilizing a tubular, microporous expanded polytetrafluoroethylene (ePTFE) membrane. The membrane itself has a 3 mm inner diameter to match standard medical tubing, keeping the fluid velocity consistent while ensuring passing bubbles make direct contact with the hydrophobic material. A membrane length between 50 mm and 100 mm provides enough contact time to clear gas from the fluid line. This membrane is enclosed by an outer housing that contains a vacuum volume. To draw the trapped air out, a compact 12V micro-diaphragm pump, such as the KNF NMP 830, is connected to this outer housing to pull a vacuum. The pressure inside the housing must be lower than the partial pressure of

dissolved nitrogen (~570 mmHg absolute) to pull the gas through the membrane pores (Yartsev 2025). However, the vacuum must stay safely above the Armstrong limit (~47 mmHg absolute at 37°C), which is the threshold where the water in blood boils and causes severe hemolysis. (Tarver 2022) Operating at 187.5 mmHg absolute, the selected pump provides rapid gas removal while providing a physical safety floor that naturally prevents the fluid from boiling.

To complement the air removal system, a bubble detection method is integrated into the fluid loop. An inline ultrasonic bubble detector is placed downstream of the membrane assembly to check for any remaining bubbles. If any air manages to pass the membrane, the sensor immediately triggers an automatic system shutdown to ensure patient safety.

4.3 SYSTEM ARCHITECTURE

To facilitate reliable and precise communication between our control software and the motor that powers the pump, we plan on using an Arduino-based control system. The Arduino will be responsible for processing inputs, executing commands, and communicating with the motor at all times.

The motor in question will be a brushless DC motor with a built-in driver and Frequency Generator (FG) signal output for feedback. The FG signal provides motor speed data via pulse frequency, which will be vital for our feedback loops. Additionally, the motor supports PWM speed control and CW/CCW direction switching via dedicated signal wires, allowing for dynamic adjustments to be made in real time if needed.

At this stage of development, we have successfully achieved the ability to run the motor continuously in both clockwise and counterclockwise directions at a desired speed. This foundational capability confirms that our hardware and software are communicating effectively and gives us a reliable base to build upon as we continue to expand the system's functionality.

Beyond the core motor system, we plan to incorporate a suite of sensors into our architecture to continuously monitor key parameters such as flow rate, air bubble detection, and any additional data points the system may require. Our current implementation already includes real-time RPM monitoring, allowing us to observe live motor speed and adjust on the fly as needed. Furthermore, we plan on incorporating the KNF NMP 830 Vacuum Pump into our architecture to rapidly remove gas while providing a physical safety floor that naturally prevents the fluid from boiling.

Looking ahead, all components will be tied together through a closed-loop feedback system, creating a fully integrated solution capable of powering the pump while precisely controlling

critical inputs such as flow rate and volume dispensed. Through this integrated architecture, our system will be able to transfuse blood in a consistent and precise manner, which will facilitate the development of an efficient, fully functional, automated pump system.

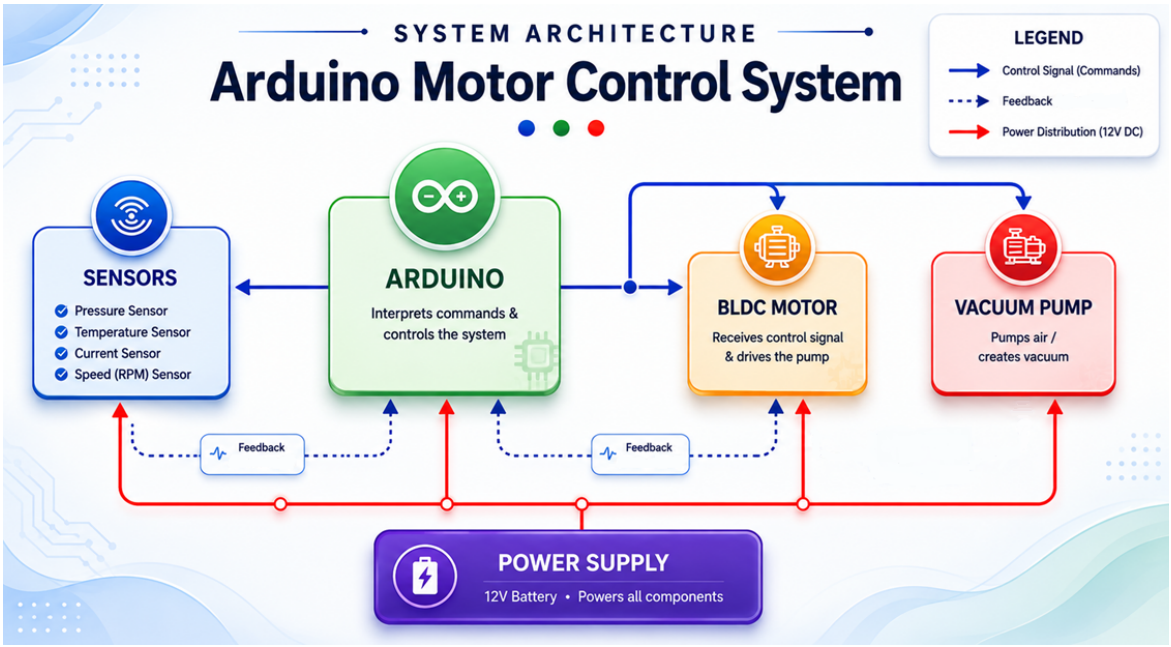

*Image generated using ChatGPT (OpenAI, 2026)

4.4 SENSORS

For detecting air bubbles in the transfusion line, the most practical and space-ready option is a clamp-on ultrasonic bubble detector, specifically the SMD A330 Bubble Sensor. This type of sensor works by sending sound waves through the tubing and measuring how they change when air is present. It is noninvasive, compact, and already widely used in medical pumps, which makes it reliable for a space environment where contamination control is critical. Ultrasonic detection is also particularly well suited to microgravity because it interrogates the full cross-section of the tubing simultaneously, meaning it does not rely on bubbles migrating to a predictable location within the tube as they would under gravity. The A330 sensor has a detection threshold of 1 µL, well below the 20 µL system safety requirement, providing comfortable margin without requiring any additional sensing hardware. An optical channel was considered but deprioritized, since optical methods require the bubble to pass through a fixed sensing region and are less reliable without buoyancy-driven phase separation.

To ensure the degassing system operates consistently, a small absolute pressure sensor is included on the vacuum reservoir. Rather than running the vacuum pump continuously, the pressure sensor allows the system to monitor the vacuum level and only activate the pump

when needed, switching it off once the target pressure is reached. This keeps power consumption low and avoids unnecessarily stressing the pump over a long operation. The NXP MPX series was selected for this role given its small size, low power draw, and straightforward integration with the Arduino.

To detect occlusions, the most sensible choice is a load-cell clamp that measures how much the tubing expands as pressure builds. When the line is blocked, the pump keeps pushing and the tubing swells slightly against the clamp, triggering an alarm. One important microgravity consideration is that the tubing must be mechanically constrained within the pump bay, since the load cell requires a fixed reaction surface to detect expansion. Motor current monitoring is included as a complementary method, since occlusions also cause the motor to work harder. In an ideal system, an inline MEMS pressure sensor would be added to help distinguish between different occlusion types and assist in troubleshooting.

For flow verification, the primary approach is motor step counting combined with a calibrated volume-per-revolution model, which gives a reliable flow estimate without additional hardware. An encoder is included to confirm the motor is turning as commanded, which is important during priming and under back-pressure conditions. Direct ultrasonic flow measurement was investigated as an upgrade option, however commercially available clamp-on ultrasonic transit-time flow sensors are expensive ($3,000+) and cannot measure speeds as low as blood infusion, making it not a currently viable solution.

For leak detection, IR optical reflectance sensors are placed near high-risk areas such as tubing connectors and near fittings. In microgravity, leaked fluid form floating droplets rather than pooling, making conductive strip detectors unreliable as a primary method since they depend on liquid bridging two conductors. IR sensors are better suited as the primary approach because they detect surface moisture films before droplets detach. Conductive strips are retained as a secondary backup.

To monitor pump health, a 3-axis MEMS accelerometer is mounted on the pump body to detect abnormal vibration patterns indicating bearing wear or rotor imbalance, compared against a baseline recorded at commissioning. For electronics health, board-mounted DS18B20 temperature sensors and Arduino ADC voltage monitoring detect overheating and power instability. Both are lightweight and straightforward to integrate into the control board.

4.5 MATERIAL SELECTION

A material trade study was conducted to determine appropriate materials for the pump structure and electronics enclosure. Candidate materials were evaluated using a weighted

decision matrix based on structural performance, manufacturability, mass efficiency, cost, and integration simplicity. Each criterion was scored on a 1–5 scale (5 = best performance).

Criteria	Weight	PLA	Polycarbonate	AL 6061	Stainless Steel
Structural Stiffness	0.25	2	3	5	5
Manufacturability	0.20	5	5	4	3
Mass efficiency	0.15	4	4	3	1
Cost	0.15	5	4	4	2
Integration simplicity	0.15	4	5	5	3
Biocompatibility relevance	0.10	3	3	4	5
Total	**1**	**3.75**	**4**	**4.25**	**3.25**

The results of the weighted matrix indicate that 6061-T6 aluminum provides the best balance of stiffness, dimensional stability, and manufacturability for the mechanically critical pump body and roller assembly. These components require consistent alignment and compression to ensure reliable peristaltic pump performance. Although aluminum introduces a modest mass increase, enclosure mass is a secondary consideration at the current system scale relative to mechanical reliability and structural stability.

A polycarbonate enclosure was selected for the electronics housing due to its low mass, ease of manufacturing, electrical insulation properties, and sufficient structural performance for non-load-bearing applications. Transparent or translucent polycarbonate also enables visual inspection of internal components during testing and troubleshooting.

While stainless steel provides excellent strength and corrosion resistance, its higher mass and increased manufacturing complexity make it less favorable for this application. PLA and other 3D printed polymers remain valuable for rapid prototyping and iterative refinement of subsystem packaging and assembly interfaces during early-stage development.

Based on the trade study, the final design adopts an aluminum pump structure with a polycarbonate electronics enclosure to balance structural performance, manufacturability, and subsystem integration requirements.

5 MASS, VOLUME AND POWER ESTIMATES

Preliminary mass, volume, and power estimates were developed to verify compliance with system-level integration constraints (<10 lbs. total mass, 512 in^3 volume, and >3 hours continuous operation). A minimum of 3 hours of continuous operation was chosen to enable

completion of a full blood-unit transfusion with operational safety margin for monitoring, flow adjustment, and contingency conditions.

5.1 POWER AND ENERGY REQUIREMENTS

Power consumption was estimated for each component from manufacturer's datasheets, with the results summarized in the table below. Component power draw is given by $P = V \times I$, and total energy required by $E = P \times t$.

Component	**Voltage**	**Typical Current**	**Avg. Power**
KNF NMP 830 Vacuum Pump	12V DC	0.25 A (15% duty)	0.45 W
Brushless DC Motor (Peristaltic)	12V DC	0.13 A	1.56 W
Ardunio Microcontroller	5V (regulated)	0.05 A	0.25 W
Sensors and Signaling Condition*	3.3-12V	~0.9A	0.51 W
Wiring & Regulator Losses (~10%)	–	–	0.21W
Total			**2.98 W**

*Includes ultrasonic bubble detector, vacuum pressure sensor, load cell amplifier, MEMS accelerometer, and temperature sensors.

The vacuum pump operates intermittently under pressure-sensor control, switching off once the target vacuum is reached and cycling back on only when reservoir pressure rises above the setpoint. At an estimated 15% duty cycle (i.e. running 15% of the time), its average power contribution is 3.0 W × 0.15 = 0.45 W. The vacuum pump and peristaltic motor account for approximately 67% of total system power, while all sensors and signal conditioning combined contribute less than 17%, confirming that sensor selection has minimal impact on battery sizing.

The minimum energy required for a three-hour mission is 2.98 W × 3 h = 8.94 Wh. A 50% margin is applied to account for estimation uncertainty and to avoid deep discharge, which degrades lithium cell cycle life, yielding a minimum required capacity of 17.88 Wh, or approximately 1490 mAh at 12 V. A 12.8 V, 3 Ah $LiFePO_4$ battery (38.4 Wh) was selected, providing a safety factor of ~2 over this minimum. $LiFePO_4$ chemistry was chosen for its inherent thermal stability, a critical consideration for energy storage aboard a crewed spacecraft and its long cycle life of approximately 2,000 charge cycles. Furthermore, the batteries offer superior mass savings, up to 3-4 times compared to lead-acid batteries.

5.2 MASS BUDGET

Component	Estimated Mass (kg)
BLDC Motor	0.075
Vacuum Pump	0.175
Arduino Microcontroller	0.2
$LiFePO_4$ Battery (3Ah)	0.23
Tubing & Connectors	0.2
ePTFE Membrane Cartridge	0.1
Metal Pump Structural Housing	0.3
Polycarbonate housing	0.7
Sensors	0.1
Total Estimated Mass	**2.08 kg (4.56 lbs.)**

The estimated total system mass is 4.56 lb., providing a significant margin relative to the 10 lb. system requirement.

5.3 VOLUME ESIMATES

The system is designed to meet a stowed volume allocation of no more than 512 in^3 (8.39 L), selected as a compact packaging constraint to support integration within spacecraft medical storage environments. This value serves as a preliminary design requirement and provides a margin for subsystem integration, routing, and detailed CAD refinement during later design stages.

The system architecture is organized into modular subsystems to support compact integration, functional isolation, and maintainability:

- Pump and actuation assembly for fluid transport and flow control
- Electronics module for sensing, control, and system regulation
- Battery system for autonomous operation and power delivery
- Disposable fluid path cartridge for sterile, replaceable blood handling

This modular configuration is intended to support integration within constrained spacecraft stowage environments while maintaining accessibility for maintenance, subsystem isolation, and ease of assembly.

6 VALIDATION AND VERIFICATION

System verification will be conducted using inspection, analysis, demonstration, and testing to confirm compliance with derived platform requirements. Verification methods were selected based on requirement type (performance, safety, environmental, and integration constraints). Testing will progress from subsystem bench validation to fully integrated system testing. The table summarizes primary verification activities.

Requirement ID	Requirement Summary	Verification Method	Validation Approach
PFR_5	Operate in microgravity (0 g)	Environmental Testing	Reduced-gravity analog testing and functional performance verification
PFR_6	Operate in partial gravity (<1 g)	Environmental Testing	Tilt-table or reduced-gravity simulation testing
PFR_14	Detect and stop flow for air ≥20 μL	Functional Testing	Controlled air bubble injection and automatic shutoff validation
PFR_16	Limit hemolysis to ≤0.8%	Laboratory Testing	Pre- and post-infusion blood sample analysis
PFR_23	Deliver 400–500 mL per transfusion	Performance Testing	Volume measurement using calibrated collection apparatus
PFR_24	Maintain 100–250 mL/hr flow rate	Performance Testing	Flow sensor validation across operating range
PFR_9	Support ≥50 transfusion cycles	Lifecycle Testing	Repeated-use endurance testing under nominal load
PFR_11	Total mass ≤10 lb	Inspection	Physical measurement using calibrated scale
PFR_18	Provide real-time monitoring and alarms	Functional Demonstration	Sensor integration and alarm response validation

Verification testing will be conducted in progressive phases, beginning with individual component validation (pump, bubble trap, sensor systems) and culminating in integrated system testing under representative operating conditions. Environmental and lifecycle testing will confirm reliability and safety prior to operational deployment.

We acknowledge that blood has an optimal working temperature range, higher than ambient. While temperature management is outside the current scope, as we assume blood has been prepared at the desired temperature prior to use, this represents a secondary consideration that could be investigated at a more advanced technology readiness level, once the primary challenge of bubble mitigation has been addressed.

7 Risk Assessment and Mitigation

A structured risk assessment was conducted to identify technical, safety, and integration risks associated with the transfusion system. Risks were evaluated qualitatively based on likelihood and consequence, and mitigation strategies were incorporated into the system design and verification plan. The table below summarizes primary project risks and corresponding mitigation approaches.

Risk Category	Risk Description	Likelihood	Impact	Mitigation Strategy
Clinical Safety	Air embolism due to sensor or shutoff failure	Low	High	Two independent ultrasonic sensors, automatic shutoff, and manual override
Clinical Safety	Excessive hemolysis from mechanical shear	Medium	High	Low-RPM gear reduction, compliant tubing selection, laboratory validation
Mechanical	Pump stall or gearbox failure	Medium	Medium	Torque margin in motor sizing, lifecycle endurance testing
Mechanical	Vacuum pump or pressure seal failure	Low	High	Dual pressure sensors with fault alarming, backup seal, and pre-use leak-down self-test
Electrical	Battery depletion during operation	Medium	High	Energy margin (>20%), real-time battery monitoring
Environmental	Performance degradation in reduced gravity	Medium	High	Reduced-gravity analog testing prior to deployment
Contamination	Breach in sterile fluid path	Low	High	Closed disposable tubing set, needle, and membrane filtration
Integration	Exceeding mass or volume constraints	Low	Medium	Preliminary mass budget with design margin

8 Cost, Schedule, and TRL Roadmap

8.1 Prototype Cost Estimate

A preliminary cost estimate was developed based on commercially available components and CNC machining of structural elements. Purdue University provides students with access

to CNC machining resources at no cost, which significantly reduces manufacturing expenses that would otherwise represent a substantial portion of the budget. The design emphasizes off-the-shelf hardware to reduce development risk and maintain affordability for laboratory-scale prototyping. Being a student-led project also gives the benefit of unpaid labor in research and developmental stages. The university is providing cost sharing funds to cover the advisor's contribution because this project is being developed as part of an academic credit course.

Component	**Estimated Cost (USD)**
BLDC Motor	$14
Vacuum Pump	$320
Arduino Microcontroller	$28
$LiFePO_4$ (3Ah)	$32
Tubing & Connectors	$100
ePTFE Membrane Cartridge	$120
Metal Pump Structural Housing	$100
Polycarbonate Housing	$100
Sensors	
Ultrasonic Bubble Detector	$250
Vacuum Pressure Sensor	$30
Load Cell	$15
Leak Detection	$20
3-Axis MEMS Accelerometer	$20
Temperature Sensors	$20
Estimated Subtotal	**$1,169**
Testing Consumables & Contingency (~20%)	$234
Total Estimated Prototype Cost	**$1403**

The estimated prototype cost reflects laboratory-scale subsystem validation and integrated system demonstration. High-cost environmental qualification activities (e.g., reduced-gravity flight campaigns) are considered future development efforts beyond the scope of initial prototype demonstration.

8.2 DEVELOPMENTAL, TEST, AND EVALUATION SCHEDULE

The proposed system is currently assessed at TRL 2-3, supported by analytical modeling and component feasibility evaluation. Advancement to flight readiness follows a phased maturation strategy aligned with NASA's Development, Test, and Evaluation (DT&E) framework.

Years 0 - 1 (TRL 3→4): Finalize detailed design and construct a laboratory prototype integrating the peristaltic pump, membrane-based bubble trap, and embedded control architecture. Subsystem validation will include flow accuracy, bubble detection performance, and power endurance testing, culminating in >3-hour continuous operation demonstration.

Years 1 - 3 (TRL 4→5): Conduct environmental testing representative of operational conditions, including vibration, thermal cycling, vacuum exposure, and reduced-gravity validation via parabolic flight. Successful demonstration advances the system to component validation in a relevant environment.

Years 3 - 5 (TRL 5→6): Develop a flight-representative engineering model incorporating structural hardening, radiation-tolerant electronics where required, and redesign of the electrical interface for compatibility with regulated spacecraft power buses. Integrated system testing under flight-like conditions advances the design to TRL 6.

Years 5 - 8 (TRL 6→7/8): Complete qualification-level vibration, shock, and thermal vacuum testing; finalize interface control documentation; perform hazard and safety certification; and integrate with a host exploration platform. This progression enables readiness for mission deployment within an Artemis-relevant timeframe.

9 VALUE PROPOSTION AND PATH FORWARD

Sustained human exploration under the Artemis program requires reliable, low-mass medical and fluid management systems capable of operating in reduced gravity without frequent maintenance or resupply. The proposed peristaltic pump and membrane-based bubble mitigation architecture directly addresses microgravity-induced air entrainment while maintaining compact size, low power consumption, and operational simplicity.

The system leverages commercially available components to enable near-term prototype validation while preserving a clear pathway toward flight qualification. With total system mass under 10 lb. and continuous power consumption below 3 W, the design aligns with spacecraft resource constraints and long-duration mission requirements.

Future maturation efforts will focus on environmental qualification, reduced-gravity validation, and redesign of the electrical interface for compatibility with regulated spacecraft power buses. Within a structured 5–8 year development horizon, the technology can advance toward flight-ready implementation supporting crew autonomy and operational resilience in lunar and deep-space environments.

APPENDIX A: CALCULATIONS

The minimum required flow rate is:

$$100\,\frac{mL}{hr} = 2.78 * 10^{-8}\,\frac{m^3}{s}$$

For 3 mm inner-diameter tubing and four rollers, the estimated displacement per revolution is approximately:

$$V_{rev} \approx 4.2 * 10^{-7}\ m^3$$

Required rotational speed:

$$N = \frac{Q}{V_{rev}} \approx 0.066\,\frac{rev}{s} \approx \ \ 4\ RPM$$

Repeat same steps for maximum required flow rate:

$$250\,\frac{mL}{hr} = 6.94 * 10^{-8}\,\frac{m^3}{s}$$

Required rotational speed:

$$N = \frac{Q}{V_{rev}} \approx 0.16\,\frac{rev}{s} \approx \ \ 10\ RPM$$

APPENDIX B: REFERENCES

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