# Technical Blueprint for a Budget-Friendly Non-Invasive Bilirubin Monitor and Analysis of a Software-Based Alternative

## Part 1: Technical Deep-Dive: A Budget-Friendly TcB-Style Reflectance Prototype

The provided curriculum for the B.Tech. in Computer Science and Engineering (Internet of Things) outlines a robust framework for developing advanced, sensor-based projects, particularly within courses such as KOT051: Sensors, Actuators and Signal Processing, KOT052: Programming and Interfacing with Microcontrollers, and the capstone KOT651: Advance Internet of Things Lab.1 While the documents do not describe an existing prototype, they provide the necessary academic foundation to design one.

This section details a feasible, budget-friendly hardware prototype for a non-invasive transcutaneous bilirubin (TcB) monitor based on the principles of diffuse reflectance spectroscopy.

### 1.1 The Biophysical Principles of Transcutaneous Bilirubinometry

#### 1.1.1 Light-Tissue Interaction: The Skin as an Optical Medium

A TcB device operates using diffuse reflectance spectroscopy (DRS).2 When light is projected onto the skin, it does not simply reflect off the surface. It enters the tissue, a complex "turbid medium," where photons undergo multiple scattering events from cellular structures, collagen, and other components. Simultaneously, photons are absorbed by specific molecules known as chromophores. A fraction of this scattered light eventually escapes the skin and is captured by a detector.5

The central engineering challenge is that the measured reflectance is a complex function of both scattering and absorption. The device's goal is to de-convolve these two factors to isolate the light absorption caused by the target chromophore: bilirubin.

#### 1.1.2 Deconstructing the Spectrum: The Four Competing Chromophores

The primary difficulty in TcB measurement is that bilirubin is not the only, or even the strongest, light absorber in the skin. The measured signal is a composite of all chromophores present. Any successful design must be able to differentiate the bilirubin signal from three main "competitors."

1. **Bilirubin:** The target molecule. It has a characteristic absorption peak in the blue region of the spectrum, around 460-476 nm.7
2. **Hemoglobin (Hb/HbO2):** Oxyhemoglobin (HbO2) and deoxyhemoglobin (Hb) are the dominant absorbers in the dermis and the most significant confounders. Their absorption spectra are complex, with a very strong Soret band (also in the blue region, overlapping with bilirubin) and two Q-bands in the 500-600 nm range.9
3. **Melanin:** The "great filter." This is the primary chromophore in the epidermis. It is a broadband absorber, with absorption that gradually decreases from the UV to the infrared region.9 Melanin concentration (i.e., skin tone) is a major source of error in TcB devices and is responsible for racial and ethnic discrepancies in measurements.12

This creates a complex system of overlapping signals. A device that *only* measures reflected light at 460 nm would be useless; it could not distinguish between a high bilirubin concentration, a high blood concentration (hemoglobin), or dark skin pigmentation (melanin). Therefore, any functional TcB device *must* use multiple wavelengths to create a system of equations that can mathematically isolate the contribution of bilirubin.

Table 1 provides a summary of the key optical properties of these competing chromophores.

| **Table 1: Key Chromophore Optical Properties** |
| --- |
| **Chromophore** |
| **Bilirubin** |
| **Oxyhemoglobin (HbO2)** |
| **Deoxyhemoglobin (Hb)** |
| **Melanin** |

#### 1.1.3 The Mathematical Foundation: Applying the Modified Beer-Lambert Law (MBLL)

The standard Beer-Lambert Law (BLL) taught in chemistry, which describes light absorption in clear, non-scattering solutions, is invalid for tissue.5 For a turbid medium like skin, a **Modified Beer-Lambert Law (MBLL)** is applied.

This model assumes that the total absorption coefficient of the skin, $μ\_a(\text{skin})$(λ), at a specific wavelength (λ) is a linear superposition of the concentrations ($C$) of each individual chromophore, multiplied by their known, constant extinction coefficients ($ε$(λ)) at that wavelength.6

The core algorithm is based on this equation:

$$μ\_a(\text{skin})(λ) = 2.303 \times [C\_{\text{HbO}\_2} \times ε\_{\text{HbO}\_2}(λ) + C\_{\text{Hb}} \times ε\_{\text{Hb}}(λ) + C\_{\text{melanin}} \times ε\_{\text{melanin}}(λ) + C\_{\text{bilirubin}} \times ε\_{\text{bilirubin}}(λ)]$$

6

This equation *is* the computational task. If the system measures the skin's absorption $μ\_a(\text{skin})$ at *at least four* different wavelengths (λ1, λ2, λ3, λ4), it generates a system of four linear equations. Since the extinction coefficients ($ε$) are known physical constants, the only unknowns are the four concentration ($C$) values. The microcontroller's job is to solve this 4x4 system of equations to find the value of $C\_{\text{bilirubin}}$.

#### 1.1.4 Commercial Design Analysis: The Two Paths to a Solution

Commercial devices use two main approaches to solve this problem. Our prototype will emulate the more budget-friendly of the two.

* Case Study 1: The Two-Wavelength / Dual-Path Approach (Dräger JM-103/105)  
  This device uses an elegant, efficient method. It measures at just two primary wavelengths: blue (450 nm) and green (550 nm).15 The 450 nm wavelength targets the bilirubin absorption peak. The 550 nm wavelength is strategically chosen because it is near an isobestic point for hemoglobin, where oxy- and deoxyhemoglobin absorb light similarly. By measuring the optical density difference between the 450 nm and 550 nm signals, the device effectively subtracts the hemoglobin background to isolate the bilirubin contribution.15 It also employs "two optical paths"—a short path to measure the epidermal melanin and a longer path to measure the dermal bilirubin/hemoglobin—to mathematically correct for skin tone.15
* Case Study 2: The Full-Spectrum Subtraction Approach (BiliChek)  
  This is a more advanced device. It uses a white-light source (like a xenon lamp) and a miniature spectrometer to measure the entire diffuse reflectance spectrum from 400-760 nm.17 It then applies the MBLL algorithm across all of these data points, performing a "mathematical subtraction" of the hemoglobin, melanin, and other factors to deliver a highly accurate bilirubin value.18 This method is very robust against variations in skin tone 17 but is prohibitively expensive for a student prototype.

**Prototype Choice:** The proposed budget-friendly prototype will emulate the **Dräger JM-105 model**, using a simplified, two-wavelength (460 nm and 550 nm) ratiometric approach. This is highly feasible for a B.Tech project.

### 1.2 Prototype Hardware Architecture and Circuit-Level Design

This design connects skills from KOT052 (Microcontrollers) and KOT051 (Sensors). The system consists of three modules: (1) Illumination, (2) Detection/Amplification, and (3) Digitization/Processing, all controlled by a central microcontroller (e.g., an Arduino Nano, or an ESP32 for IoT connectivity).

#### 1.2.1 The Illumination Module (The "Light Source")

This module must generate a highly stable, non-flickering light output. Because the final measurement is a *ratio* of two signals, any flicker or thermal drift in the LED brightness will corrupt the data. Powering the LEDs with a simple resistor is unacceptable for a medical-grade measurement.

* **Component Selection:** Two high-intensity, narrow-band LEDs:
  1. **~460 nm** (Blue): To target the bilirubin absorption peak.15
  2. **~550 nm** (Green): To provide the hemoglobin reference signal.15
* Circuit Design: Op-Amp Constant Current Driver  
  A simple and reliable constant current source can be built using a basic LM358 op-amp (or similar) and an NPN transistor.23
  1. The microcontroller (a KOT052 skill) outputs a PWM signal. This is passed through a simple RC low-pass filter to create a stable, analog reference voltage ($V\_{\text{ref}}$).23
  2. $V\_{\text{ref}}$ is fed into the op-amp's non-inverting input.
  3. The op-amp's output drives the base of the transistor, which controls the LED. A low-value sense resistor ($R\_{\text{sense}}$) is placed in series with the LED, and the voltage across it is fed back to the op-amp's inverting input.
  4. The op-amp's feedback loop will force the voltage at $R\_{\text{sense}}$ to be equal to $V\_{\text{ref}}$. By Ohm's Law, the current through the LED is *locked* at $I\_{\text{LED}} = V\_{\text{ref}} / R\_{\text{sense}}$. This current remains stable regardless of temperature changes in the LED or fluctuations in the main power supply.

#### 1.2.2 The Detection and Amplification Module (The "Light Sensor")

The diffuse light reflected from skin is extremely weak. A simple phototransistor or light-dependent resistor (LDR) from a basic lab kit 1 lacks the sensitivity and speed required. A bare photodiode (like a BPX61) produces a current in the pico-amp to nano-amp range, which is far too small to be read by a microcontroller and is highly susceptible to noise.24

* Component Analysis: The OPT101 Monolithic Photodiode  
  The ideal component for this "budget" prototype is the Texas Instruments OPT101.26 This component is a "monolithic photodiode with on-chip transimpedance amplifier".25
  + It integrates the photodiode *and* the critical transimpedance amplifier (TIA) onto a single chip. This design "eliminates the problems commonly encountered in discrete designs such as leakage current errors, noise pick-up, and gain peaking due to stray capacitance".25
  + It has a large 0.09 x 0.09 inch photodiode, an internal 1MΩ feedback resistor, and high responsivity (0.45 A/W at 650nm).27
  + Crucially, it takes in light and outputs a clean, linear, high-level *voltage* that can be directly read by an ADC.27 It is specifically designed for medical and laboratory instrumentation.26
* **Circuit Design:** The circuit for the OPT101 is simple: connect power (e.g., 5V), ground, and read the voltage from the output pin.30

#### 1.2.3 The Digitization and Processing Module (The "Brain")

The voltage signal from the OPT101, while clean, will be small. The *difference* in reflected light between the 460 nm and 550 nm wavelengths will be even smaller. The standard 10-bit ADC on an Arduino (providing 1024 steps) is too coarse and will not provide the resolution needed to detect small changes in bilirubin.

* Component Analysis: The ADS1115 16-bit ADC  
  The solution is a low-cost, high-precision ADC like the Texas Instruments ADS1115.33
  + It provides 16 bits of resolution (65,536 steps) and communicates via I2C, a protocol taught in the KOT052 course.1
  + Its most important feature is the built-in **Programmable Gain Amplifier (PGA)**.33 If the reflected light signal from the OPT101 is only in the 0-200 mV range, the student can *program* the PGA to set its full-scale input range to ±256 mV. This "zooms in" the entire 16 bits of resolution onto that tiny signal, providing the high precision necessary for a medical screening device.
* **Interfacing:** The connection is straightforward: OPT101 V\_out -> ADS1115 A0\_in. The ADS1115 SCL/SDA pins connect to the microcontroller's I2C bus.

#### 1.2.4 Complete Prototype Schematics and Bill of Materials (BOM)

The modules are integrated as follows: The microcontroller sequentially activates the constant-current driver for the 460 nm LED, takes a high-resolution reading from the ADS1115, deactivates the first LED, activates the 550 nm LED, and takes a second reading.

| **Table 2: Prototype Hardware Bill of Materials (BOM)** |
| --- |
| **Module** |
| **Processing** |
| **Digitization** |
| **Detection** |
| **Illumination** |
|  |
|  |
|  |
| **Passives** |

### 1.3 Firmware and Algorithmic Implementation

#### 1.3.1 Data Acquisition and Noise Reduction

The firmware's core loop must be carefully timed to prevent optical crosstalk. This logic is a direct application of the signal processing concepts from KOT051.1

*Pseudocode for Data Acquisition:*

1. configure\_ADC\_gain(ADS1115\_PGA\_GAIN\_16) // Set PGA to ±0.256V
2. LED\_460\_OFF()
3. LED\_550\_OFF()
4. V\_dark = average\_ADC\_read(100) // Measure system dark noise 25
5. LED\_460\_ON()
6. delay(10) // Allow sensor to stabilize
7. V\_460 = average\_ADC\_read(100)
8. LED\_460\_OFF()
9. LED\_550\_ON()
10. delay(10) // Allow sensor to stabilize
11. V\_550 = average\_ADC\_read(100)
12. LED\_550\_OFF()

The average\_ADC\_read(100) function reads the ADS1115 (which can sample at up to 860 samples per second 33) 100 times and returns the average, significantly reducing random noise.

#### 1.3.2 Implementing the Two-Wavelength Bilirubin Algorithm

The true reflectance signals are calculated by subtracting the dark noise.

1. Reflectance\_460 = V\_460 - V\_dark
2. Reflectance\_550 = V\_550 - V\_dark

The core calculation, based on the Dräger method, is the logarithmic ratio (Optical Density difference) between the reference and the target:

Bilirubin\_Metric = log10(Reflectance\_550 / Reflectance\_460)

This unitless Bilirubin\_Metric is the raw output of the device. It will be linearly correlated with the actual bilirubin concentration.

#### 1.3.3 IoT Integration

As required by KOT651, this project should solve a societal problem using IoT.1 The microcontroller (if an ESP32 is used) can connect to Wi-Fi (a KOT052 skill 1) and send the final calculated Bilirubin\_Metric (or the final mg/dL value after calibration) to a cloud server. This could be a simple PHP/MySQL backend, as referenced in the KOT052 syllabus 1, or an MQTT broker as covered in KOT601.1 This enables remote monitoring of neonatal health data by clinicians.

### 1.4 Critical Validation: Fabricating Low-Cost Tissue Phantoms

#### 1.4.1 The Necessity of Calibration

This prototype, as-is, is scientifically useless. It *cannot* be tested on a human; it is unethical and provides no meaningful data without a "ground truth." The project *must* be calibrated using "tissue phantoms"—synthetic materials that mimic the optical absorption and scattering properties of human skin.2

The Bilirubin\_Metric is just a number. Calibration is the process of creating a function f(x) where TSB\_mg\_dL = f(Bilirubin\_Metric). This function is found by testing the device on phantoms of *known* concentrations.

#### 1.4.2 Protocol 1: Gelatin-Intralipid-Bilirubin (High-Fidelity) Phantom

This protocol creates a high-fidelity phantom.

* **Base:** 5% Gelatin (e.g., Sigma-Aldrich G1890).38
* **Scatterer:** 1% Intralipid (a lipid emulsion) to mimic skin scattering.38
* **Absorber:** Bilirubin powder (e.g., Sigma-Aldrich B4126).38
* **Preparation:** The key difficulty is that bilirubin powder is not water-soluble. It must first be dissolved in **dimethyl sulfoxide (DMSO)**, then diluted with water, and finally mixed with the liquid gelatin and intralipid solution before it congeals.38 A student would create a 5-point calibration set (e.g., phantoms with 0, 5, 10, 15, and 20 mg/dL of bilirubin).

#### 1.4.3 Protocol 2: Agarose-Coffee-Tartrazine (Budget-Friendly) Phantom

The high-fidelity protocol uses expensive and hazardous chemicals (DMSO). A cheaper, safer, and highly effective alternative exists.39

* **Base:** Agarose or gelatin.39
* **Scatterer:** Lipid emulsion (e.g., Intralipid).39
* **Melanin Substitute:** Brewed **coffee solution**. The melanoidin in coffee provides a broadband absorption spectrum similar to melanin.39
* **Bilirubin Substitute:** **Tartrazine** (common yellow food dye). It can be used as a stable and safe substitute for bilirubin.39

This is the recommended path for the B.Tech project. The student would create a "base skin" phantom (agarose + intralipid + coffee) and then create a 5-point calibration set by adding 5 known, precise concentrations of tartrazine.

#### 1.4.4 Building the Calibration Curve

1. The student uses the completed prototype to measure the Bilirubin\_Metric for each of the 5 phantoms.
2. The data is plotted (e.g., Bilirubin\_Metric on the y-axis vs. Phantom\_Concentration on the x-axis).
3. A linear regression is applied to find the best-fit line, $y = mx + b$.
4. This formula is then programmed into the microcontroller. The device can now instantly convert its raw Bilirubin\_Metric into a clinically relevant mg/dL concentration, completing the project.

## Part 2: Alternative Prototype: Smartphone-Based Scleral Icterus Detection

This alternative prototype shifts the engineering challenge from complex analog hardware to sophisticated software. It is a computer-vision and machine-learning-centric project, making it an excellent fit for a student taking the KCS055 Machine Learning Techniques elective.1

### 2.1 Rationale for an Alternative: Sclera vs. Skin

#### 2.1.1 Bypassing the Melanin Confounder

The single greatest challenge for the skin-based TcB prototype is correcting for melanin.12 This correction is complex and a major source of error across different skin tones.

The alternative approach is to image the **sclera** (the white of the eye). The sclera contains *no melanin*.40 It is, therefore, an "optically superior" 40 and "race-agnostic" 43 measurement site. Bilirubin has a high affinity for the elastin in the sclera, so jaundice (icterus) is often visible there first. This elegantly bypasses the main biophysical confounder, making bilirubin the primary coloring agent to be measured.42

#### 2.1.2 Clinical Performance and Accessibility Analysis

This software-based approach leverages the ubiquitous nature of smartphones.41 Commercial TcB devices are expensive (costing thousands of dollars), making them inaccessible in low-resource settings.45 A smartphone app is "an order of magnitude cheaper" and more accessible.42

While the accuracy of a TcB device is currently higher, smartphone apps have been shown to be "comparable" and highly effective as *screening* tools. A validation study directly comparing the Dräger JM-105 TcB device to a scleral-based app (neoSCB) provides a clear performance baseline.46

| **Table 3: Diagnostic Performance Comparison: JM-105 (TcB) vs. neoSCB (Scleral App)** |
| --- |
| **Metric** |
| **Correlation with TSB ($r$)** |
| **Sensitivity (for TSB > 14.6 mg/dL)** |
| **Specificity (for TSB > 14.6 mg/dL)** |

This data shows that while the TcB device has a stronger correlation and specificity, the smartphone app's **sensitivity is excellent (94%)**.46 This makes it an ideal screening tool, as it is highly effective at identifying infants who *need* a follow-up blood test, which is the primary goal.47

### 2.2 The Computer Vision and Machine Learning Pipeline

This project involves a four-step software pipeline to process an image of an eye and return a bilirubin estimate.

#### 2.2.1 Step 1: Image Acquisition and Standardization

The new engineering challenge is no longer analog noise, but *photometric variability*. The color of an object in a photo is corrupted by ambient lighting (e.g., a warm tungsten bulb vs. cool daylight) and differences between camera sensors.44

* Method A: The Accessory-Based Approach (BiliScreen-style)  
  This method, used by the BiliScreen app, controls the environment physically.50 It uses two low-cost accessories:
  1. A **3D-printed box** (similar to Google Cardboard) that blocks all ambient light and uses the phone's flash for consistent, uniform illumination.43
  2. A **color calibration card** placed in the frame. The software finds the known color patches on the card, compares their measured RGB values to their *true* values, and generates a color-correction matrix to fix the entire image.49
* Method B: The Accessory-Free Approach (neoSCB-style)  
  This more advanced software solution, used by the neoSCB app, requires no accessories.44 It uses a technique called "ambient subtraction."
  1. The app uses the front-facing camera to capture a *pair* of images in rapid succession.
  2. Image 1 (Screen OFF): Captures the sclera illuminated *only* by the ambient room light.
  3. Image 2 (Screen ON): The screen flashes white, acting as a "flash." This image captures the sclera illuminated by *both* the screen and the ambient light.
  4. The algorithm then computes: True\_Sclera\_Color = Image 2 - Image 1.  
     This process mathematically subtracts the ambient light contribution, isolating the true, device-illuminated scleral color.44 This is the recommended approach for a CS-focused project.

#### 2.2.2 Step 2: Sclera Segmentation

Once a color-corrected image is obtained, the algorithm must automatically find *only* the sclera pixels, ignoring the iris, pupil, and surrounding skin.43

* **Traditional CV Methods:** Techniques like color-space thresholding (e.g., in HSV or L\*a\*b\* color spaces) combined with contour-finding (cv.findContours) 60 or watershed algorithms 61 can be used. However, these methods are "brittle" and often fail due to shadows, reflections (glares), and off-angle gazes.62
* Deep Learning Method: The U-Net Architecture  
  This is the modern, robust solution and a perfect KCS055 project.63 The U-Net is a convolutional neural network architecture specifically designed for precise biomedical image segmentation.63
  + It features a symmetrical **encoder** (down-sampling path) that uses convolutional layers to extract *what* is in the image, and a **decoder** (up-sampling path) that uses transposed convolutions to reconstruct the image and pinpoint *where* the features are.63
  + "Skip connections" that link the encoder layers directly to the decoder layers allow the network to combine deep, semantic feature information with shallow, high-resolution location information.
  + The output is a binary "mask" of the eye image, where pixels belonging to the sclera are '1' and all others are '0'.63

#### 2.2.3 Step 3: Colorimetric Feature Extraction

After the sclera pixels are isolated by the U-Net mask, their average color must be quantified. Simply averaging the RGB values is incorrect, as RGB is a device-dependent color space.64

* The Solution: Conversion to CIE Color Space  
  The algorithm must convert the RGB values of the sclera pixels into a device-independent and perceptually uniform color space, such as CIE XYZ or CIE L\*a\*b\*.40 This ensures that the "yellowness" value is consistent across different phone models.
* Quantifying Yellowness: The Jaundice Eye Color Index (JECI)  
  Once in the CIE XYZ space, the yellowness can be quantified using the Jaundice Eye Color Index (JECI).40 This metric is defined using the $x$, $y$, and $z$ chromaticity coordinates:  
    
  $$JECI = z\_{D65} - z = (z\_{D65} - 1) + x + y$$  
    
  (where $z\_{D65}$ is a constant based on the D65 standard illuminant).40  
  This formula effectively distills all the color data from the sclera mask into a single, scientifically valid "yellowness" score.

#### 2.2.4 Step 4: Regression Modeling

The final step is to build a machine learning model that maps the input feature (the JECI score) to an output prediction (the TSB (mg/dL) value).

* A simple linear regression model might be sufficient, as studies show a strong linear correlation between scleral yellowness and TSB.40
* The primary challenge here is data. To train this model, the student would need access to a "ground truth" clinical dataset of scleral images paired with their corresponding blood-test TSB values.46 For the scope of a B.Tech project, a student could implement the full pipeline (Steps 1-3) and validate their JECI calculations against the published data and charts in the source research papers.40

## Part 3: Project Analysis and Recommendations

### 3.1 Comparative Feasibility for a B.Tech Project

A B.Tech (IoT) student has two excellent, high-impact, and feasible paths for this project, each testing a different set of skills from the curriculum.1

* **Path 1 (TcB Reflectance Prototype):**
  + **Focus:** Hardware-Centric, Biophotonics, Analog Electronics.
  + **Skills Tested:** Analog circuit design (op-amps, constant current), sensor interfacing (I2C), applied signal processing (noise reduction, averaging), and basic chemistry/materials science (phantom fabrication).
  + **Curriculum Alignment:** A perfect capstone for KOT051 (Sensors, Signal Processing), KOT052 (Microcontrollers, I2C, SPI), and KOT551 (Pi, Sensors).1
  + **Pros:** Results in a tangible IoT device. High potential for accuracy, approaching the r=0.93 correlation of commercial devices.46
  + **Cons:** Extremely challenging. Highly susceptible to analog noise, and success depends on careful physical fabrication (probe-skin contact, optical alignment, and phantom calibration).
* **Path 2 (Scleral Icterus App):**
  + **Focus:** Software-Centric, Computer Vision, Machine Learning.
  + **Skills Tested:** Deep Learning (TensorFlow/PyTorch, U-Net), computer vision (OpenCV), data science (color space transforms, regression), and mobile/web app development.
  + **Curriculum Alignment:** A perfect capstone for KCS055 (Machine Learning Techniques), KCS602 (Web Technology, for the server backend), and KOT052 (Python programming).1
  + **Pros:** Leverages core CS skills. Elegantly bypasses the primary biophysical confounder (melanin). Enormous potential for societal impact due to its accessibility and low cost.42
  + **Cons:** Requires a "ground truth" clinical dataset to train the final regression model. The segmentation and color-correction algorithms are non-trivial to implement correctly.

### 3.2 Expert Recommendations for Prototyping and Future Work

For a B.Tech (IoT) student, **Path 2 (The Scleral App)** is the strongest recommendation. Its primary challenges are algorithmic, which directly aligns with the computer science and machine learning (KCS) components of the curriculum.1 The hardware (a smartphone) is a given, allowing the student to focus on the complex software, data processing, and ML pipeline.

However, **Path 1 (The TCB Prototype)** is an exceptional project for a high-achieving student or team, especially one with a strong interest in biophysics and hardware engineering. It perfectly fulfills the KOT651 lab's goal of "Solving Societal problems with the help of IOT".1

Regardless of the path chosen, the project should be integrated into a full IoT ecosystem as required by the curriculum. The final bilirubin value, whether from the TcB device or the scleral app, should be securely transmitted (e.g., via MQTT or a REST API) to a cloud database (as covered in KOT052 1). This data can then be displayed on a web-based dashboard for remote clinical monitoring, completing the "Internet of Things" loop.

A final, critical admonition: neither of these prototypes is a medical device. They are engineering prototypes for *screening* only. Any attempt to use them in a clinical setting would require rigorous validation, ethics board (IRB) approval, and regulatory compliance.48

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